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Biomechanical Analysis of the Lower Extremities during Cutting Maneuvers with and without an Ankle Brace

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Biomechanical Analysis of the Lower Extremities during Cutting Maneuvers with and without an Ankle Brace

by

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Abstract

Biomechanical Analysis of the Lower Extremities during Cutting Maneuvers with and without an Ankle Brace

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The ability to move is an important component for quality of life. Injury can hamper this component reducing the quality of life. The time needed to recover from injury can have economic, psychological, and emotional impacts. Therefore it is important to determine ways to prevent injuries whenever possible. One common injury occurring to many active people is an inversion ankle sprain. Unfortunately after such an injury the likelihood it may occur again increases due to the decreased ankle stability. To help prevent ankle sprains, individuals can wear external ankle supports such as ankle tape or ankle braces. The latter of the two is reusable, does not degrade with activity, and reduces the risk of ankle injuries. However, with the reduction of the ankle's range of motion there may be an alteration in the kinetic chain dynamics of the lower extremities, which may have consequences including risk of injury at other joints, including the knee and hip. Females have a four to six fold incidence rate of ACL injuries when compared to men. Additionally 60% of ACL injuries occur during noncontact situations. This study was designed to examine whether the kinetics of the knee and hip, and the angular kinematics of the knee are affected in a female population when wearing an ankle brace during a sidestep cutting maneuver. Sixteen healthy and recreationally active females between the ages of 18-40 performed a number of dynamic movement tasks using of their preferred leg. A total of 24 experimental trials were completed over two sessions (12 in session one and 12 in session two) with half of each sessions trials being straight run throughs and the other half being sidestep cutting maneuvers. Half of the participants wore a brace during their first session (no brace for their second) and the other half did not wear a brace during their first session (brace for their second). The results showed no difference in almost all dependent variables except for knee varus moment between sessions one and two with a p-value less than 0.05. These findings suggest a learning effect occurred and the participants altered their valgus moments without altering knee angular kinematics. Wearing an ankle brace did not increase the injury risk factors for the knee in this task.

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Chapter 1: Introduction

Physical activity for individuals of all ages is advised to improve or maintain lifetime health. There are numerous benefits to being physically active which can include aerobic fitness, stronger bones, strength gains, and a better general wellbeing. However, with vigorous activity there is a risk of injury occurring during a bout of physical activity. Injuries can have a personal economic cost, take time away from exercise resulting in reduced wellness, and decrease athletic performance, therefore decreasing an individual's overall quality of life for a period of time. Inversion ankle sprains are among the most common musculoskeletal injuries.[6] Furthermore, once an individual has sprained his or her ankle there is a higher risk of injury reoccurrence due to decreased joint stability. A popular trend in treating ankle injuries is to use external ankle supports which can reduce the risk of injury or re-injury.[1-3] A commonly used ankle support is a lace-up plus strap ankle brace. However, while an ankle brace will provide support for the ankle, it is not well known whether this stabilizing or stiffening of the ankle has any effect on the knee during dynamic locomotion tasks.[4,5] Knee injuries are also commonly occurring injuries, and in a survey of injury rates among collegiate athletes from 2004-2005 through 2012-2013, anterior cruciate ligament (ACL) injury rates rose in men and women's sports.[26,6] Most ACL injuries are noncontact related (60% women and 59% men), and they are sustained by women at higher rates than men in soccer, basketball, and lacrosse.[6] These frequencies of significant ankle and knee injuries raise concerns about what causes them and whether there is a relationship between them. In particular, there is reason to be concerned about a possible link between restricted ankle movement and knee loads that results in efforts intended to protect the ankle actually hurting the knee.

Chapter 2: Background

Several studies have reported ankle bracing and taping to be equal in their effectiveness to restrict ankle motion in the frontal (eversion/inversion) and sagittal (plantar/dorsiflexion) planes.[7-9] This restriction of the ankle's ability to invert/evert is beneficial in stabilizing an injured, weakened ankle or in reducing the likelihood for an ankle injury to occur. However, limiting the range of motion of one joint in a kinetic chain such as the lower extremities may impose a greater demand on the other joints. If overall performance does not change when the range of motion at the ankle is restricted by an external ankle support, a new movement pattern with altered motion and forces at other joints in the kinetic chain must compensate for the decreased motion at the ankle.[27] Several previous studies have examined ankle bracing and its effects on the knee in a variety of tasks, with conflicting results. Analyses of knee moments during a landing or squatting task revealed an increased knee injury risk with ankle taping due to an increased external knee torque or medial knee displacement.[11,13] Others have shown when the ankle is taped there is greater knee flexion during landing, which is associated with decreased knee injury risk due to reduced leg stiffness.[12,14] These results are similar to those reported in running and sidestepping tasks evaluated in semipro rugby players that showed protective benefits to the knee by reducing peak varus moments when the ankle was taped.[5] In evaluating these studies of the effects of increased ankle support, it is important to keep in mind that while ankle taping has been used in such studies, it has a major limitation that is the tape weakens when stressed, resulting in degradation of support occurring during prolonged exercise, with maximal losses of support occurring in the first 20 minutes from the start of the exercise.[15,16]

In cutting tasks men and women often have different motor program strategies, yet most studies have evaluated only men with ankle taping, excluding a vast population.[17-20] Women have not been well represented in the literature in this area and in particular have not been studied while performing cutting tasks with an external ankle support. This is an important omission, as women experience up to four times more ACL injury occurrences than men.[21] Thus it is possible that ankle bracing might predispose female athletes to greater risk of ACL injuries, which will then require weeks to heal and months to come back fully to pre-injury performance.

This study compared the biomechanical actions of the knee with and without an ankle brace during a sidestep cutting task during running performed by recreationally active female college students. External ankle supports have been shown to reduce the range of motion in the frontal and sagittal planes, thus decreasing the net joint moments the muscles across the ankle can actively provide. It was hypothesized that ankle bracing would increase the knee valgus moment, knee extension moment, and knee external rotation moment, hip extension moment, external rotation and adduction moment while decreasing the plantar flexion moment and inversion moment at the ankle. Due to the change in kinetics, it was also hypothesized that kinematics would also be altered while wearing an ankle brace. In particular the knee joint would experience more valgus, extension, and external rotation during trials with an ankle brace when compared to trials without a brace.

Chapter 3: Methods

3.1 Participants

We recruited 16 healthy, recreationally active female volunteers (26.4±5.3 yrs, 166.1± 4.2 cm, 63.7± 9.7 kg) with no severe leg injuries within two years of participation that resulted in a fracture or any surgery performed on rigid tissues in the lower extremities within three years of participation. Participants were under no physical restrictions from a primary care giver and were between 18 and 40 years of age.

When studying musculoskeletal function in female populations, it is important to consider the menstrual cycle, due to the fluctuation of hormones and the effect they may have on the body. In particular, increasing amounts estrogen cause decreased ACL fibroblast reproduction and type I procollagen synthesis , possibly causing a weakness in ACL strength.[28] Several evaluations of the different phases of the menstrual cycle in a cutting and jump-stop landing revealed a lesser knee joint laxity during the follicular phase when compared to the other two phases.[22-24] Increases in joint laxity have been associated with increased levels of estrogen and may be helpful in explaining why females are at greater risk for ACL injury.[22,24] All data collection sessions were completed within two weeks from the start of the menstrual cycle regardless of whether a participant was using a form of birth control. It should be noted that one participant had an absence of a menstrual cycle due to excessive amounts of physical activity. However, since the objective of this time restriction was to have data collection during a period of low estrogen this participant was included.

Emails were sent to all individuals who volunteered containing a questionnaire with exclusionary criteria and to ensure the sessions were scheduled and completed within the appropriate time.

This study was approved by the University of Texas at Austin Institutional Review Board and informed consent was obtained from all participants.

3.2 Task and Procedures

There were two tasks for the participants to complete while running across a level floor and stepping on a force plate: a sidestep cut, and a straight run-through. A sidestep cut is performed by moving to the side opposite the planted foot (if the left foot plants, the participant moves to the right). All participants completed two sessions of 12 trials that took place on separate days within two weeks from the start of the participants' menstrual cycle. Each session was performed either with or without a brace; participants who completed the first session with a brace performed the second session without a brace and vice versa. The order in which the participants completed the two conditions was assigned in a pseudorandom fashion using a random number generator in Microsoft Excel 2011 so that half the participants performed the braced condition in their first session, and the other half performed the unbraced condition in their first session.

A trial was deemed successful if the participant met an approach speed deemed acceptable by the investigator and made contact wholly with the force plate with her preferred foot from which to make a sidestepping cut. To determine the preferred sidestepping foot, participants performed sidesteps with both their right and left legs. The foot that the participant thought was more comfortable from which to perform the sidestep was chosen as the preferred sidestepping foot. Tasks were organized pseudorandomly in each session so that each task was performed six times. The cutting motions were required to be performed from the center of the force place at an angle between 45 and 75-degrees in the anterior direction. All movements were performed with the preferred foot planting on a Bertec force plate mounted flush with the floor surface. A pressure pad was placed approximately 1.5 m before the force plate on the 10 m x 1 m runway to detect the non-preferred foot contact so that running speed could be monitored, assuring that it was consistent for all trials. Participants were instructed not to focus on contacting the pressure pad for all tasks and rather to focus on the force plate located six meters from the start of the runway. Two strips of blue tape were placed to the right and left sides of the running approach path. Beyond the force plate similar markings indicating the cutting lanes (from 45 degrees to 75 degrees from an extension of the approach path), centered on the force. Only the preferred sidestepping ankle was braced. Lace-up and strap BCG ankle braces (BCG, TX) for shoe sizes 8-11 or 11-14 were used. Application of the brace was supervised by the lead investigator.

After consent was obtained and the health questionnaire was completed, anthropometric measures of each participant were taken to allow inverse dynamic calculation of joint moments. Leg length was measured from ASIS to medial malleolus, knee width at the lateral and medial points of the epicondyles of the tibia, and ankle width from the lateral malleolus to medial malleolus. After these measures were recorded, each participant performed a 10-minute warm-up at 75 W on a stationary bike. A task familiarization followed that determined the participant's preferred sidestepping foot. At the start of familiarization the participant performed multiple straight runs until she was running at an estimated speed of 5 m/s. Once she could consistently produce the desired velocity, the participant's starting point was adjusted so that she consistently hit the force plate with her entire foot. After the starting point was set each participant practiced sidesteps with both right and left feet to determine her preferred sidestepping foot. Then each participant practiced until she had performed at least two successful sidesteps, with success being defined the same as during the data collection trials. If the participant was to wear a brace for a session, the familiarization protocol was performed again with the brace on. Sixteen reflective markers were placed on the lower limbs in accordance to the Vicon Nexus Plug-in-Gait manual. These locations were on both the right and left lower limbs: head of second metatarsal, lateral malleolus, heel, anterior mid-shank, lateral epicondyle of the tibia, lateral mid-thigh, ASIS and PSIS. Data collection then proceeded with the participant being informed of the specific task before starting each trial, asking if she was ready, waiting 2 sec to allow the investigator to start data capture and watch the task, then instructing her to perform the stated task.

3.3 Data Acquisition

All data collection sessions were completed in the Developmental Motor & Cognition Lab, located on the 5th floor of Bellmont Hall at The University of Texas at Austin campus. Motion data were captured with Vicon Nexus 1.8.5 (Vicon Motion Systems, UK), recording at 120 Hz. Ground reaction forces were recorded at 1200 Hz with a 6 dof Bertec force platform (Bertec, Ohio). Knee width and ankle widths were measured using anthropometric calipers (Lafayette Instrument Company, IN). Leg lengths were measured using a Komelan measuring tape. Height and weight were measured using a clinical scale.

3.4 Data Analysis

Knee angles, knee moments and hip moments were evaluated at initial contact (IC), at the time of maximum ground reaction force (max GRF), and at maximum knee flexion (MKF) for the order conditions session one (S1) and session two (S2) and the brace conditions braced (B) and no brace (NB). Straight run through trials were not analyzed but were used as catch trials. Data were processed and exported using Vicon Nexus 1.8.5, and a custom Matlab script (Matlab 2017b, Mathworks MA) was created to aid in analysis of the data. Significance tests of the kinetic and kinematic data were completed with a 2x2 repeated measures ANOVA ($p < 0.05$) via SPSS version 25. Bonferroni adjusted t-tests were used for post-hoc analysis if indicated.

Chapter 4: Results

4.1 Moments and Angles at Initial Contact

On average, trials were performed at 3.65 ± 0.23 m/s in session one and at 3.69±0.23 m/s in session two with no significant difference between sessions (t=0.5438, p=0.5906). At the time of IC there was not a significant interaction effect when comparing the brace*order interaction for the hip and knee moments. For hip moments there were no statistical differences between S1 and S2 hip flexion/extension moments (S1: 2.31±0.70 Nm; S2: 2.53±0.73 Nm; p>0.05; see figure 1), hip adduction/abduction moments (S1: -0.21±0.64 Nm; S2: -0.79±0.34 Nm; p>0.05; see Figure 1), and internal/external rotation moments (S1: -0.043±0.05 Nm; S2: 0.040±0.06 Nm; p>0.05; see Figure 1).

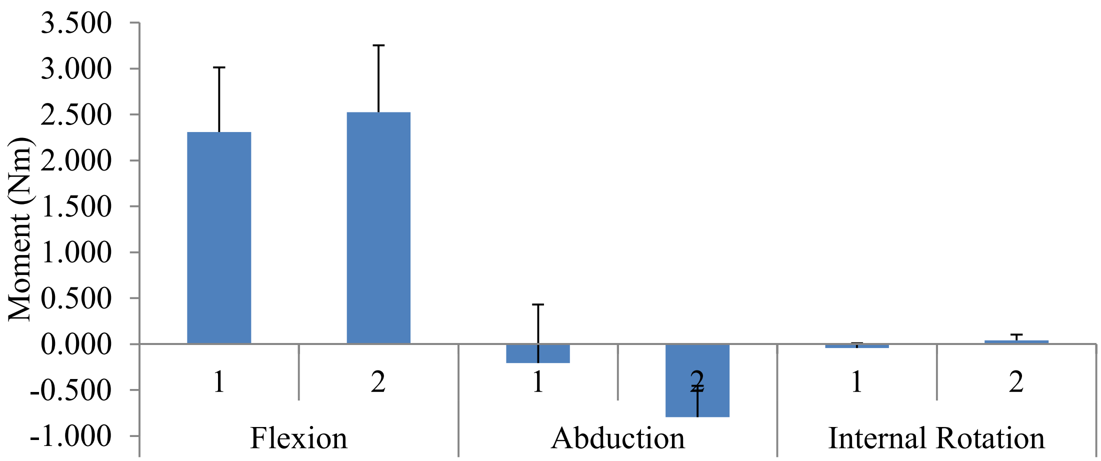

Figure 1. S1 vs S2 averages at IC for the hip joint moments; no significant

In addition, data from the B and NB conditions did not differ significantly for the hip flexion/extension moments (B: 2.37±0.73 Nm; NB: 2.46±0.67 Nm; see Figure 2), hip adduction/abduction moments (B: -0.39±0.71 Nm; NB: -0.61±0.25 Nm; see Figure 2), and the hip internal/external rotation moments (B: 0.00±0.65 Nm; NB: 0.00±0.039 Nm; see Figure 2).

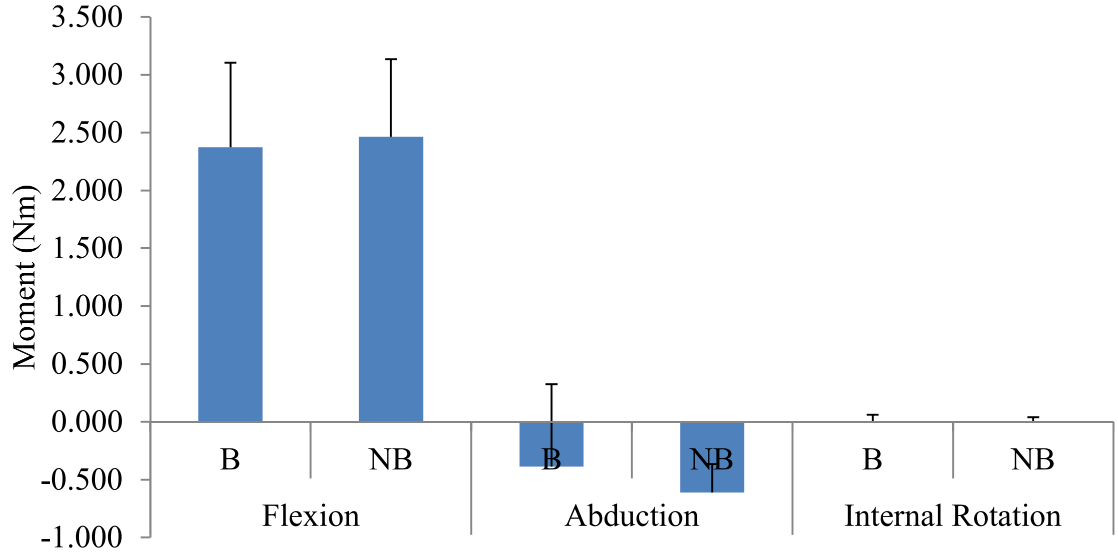

Figure 2. Brace vs no brace condition averages at IC for hip moments; no

Comparison of the knee moments from session one to session two yielded no statistical differences for the knee flexion/extension moments (S1: -0.99+0.34 Nm; S2: -1.00±0.40 Nm; $p>0.05$; see Figure 3), knee valgus/varus moments (S1: -0.095±0.41 Nm; S2: -0.23±0.36 Nm; $p>0.05$; see Figure 3), and knee internal/external moments (S1: -0.012±0.095 Nm; S2: -0.052±0.055 Nm; $p>0.05$; see Figure 3).

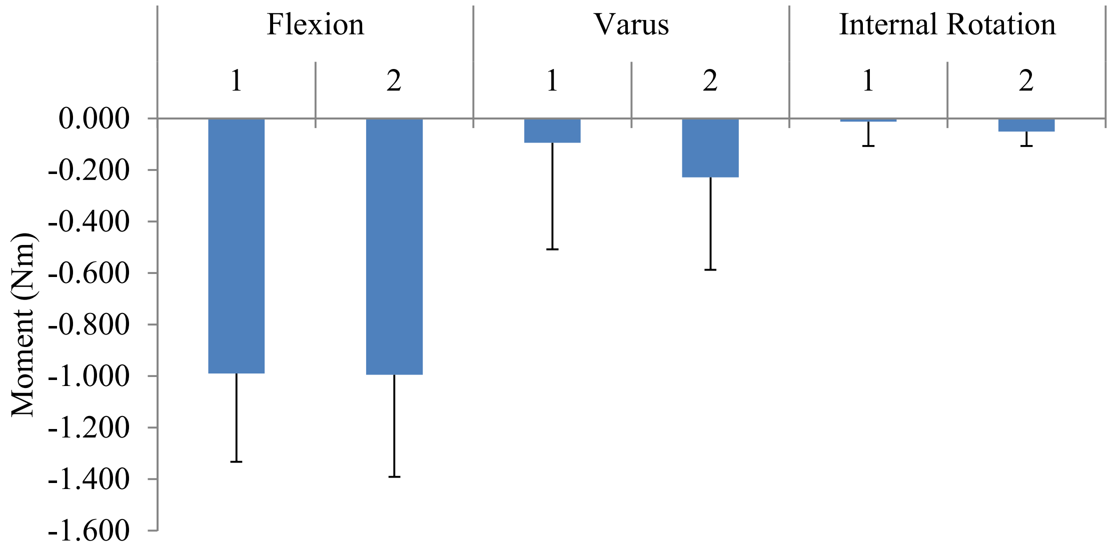

Figure 3. S1 vs S2 averages at IC for the knee joint moments; no significant

Comparison of the conditions B and NB also yielded no statistical differences for the knee flexion/extension moments (B: -0.98 $\pm$0.36 Nm; NB: -1.00$\pm$0.38 Nm; $p>0.05$; see Figure 4), knee varus/valgus moments (B: -0.18$\pm$0.48 Nm; NB: -0.14$\pm$0.29 Nm; $p>0.05$; see Figure 4), and the knee internal/external rotation moments (B: -0.042$\pm$0.11 Nm; NB: -0.022$\pm$0.044 Nm; $p>0.05$; see Figure 4).

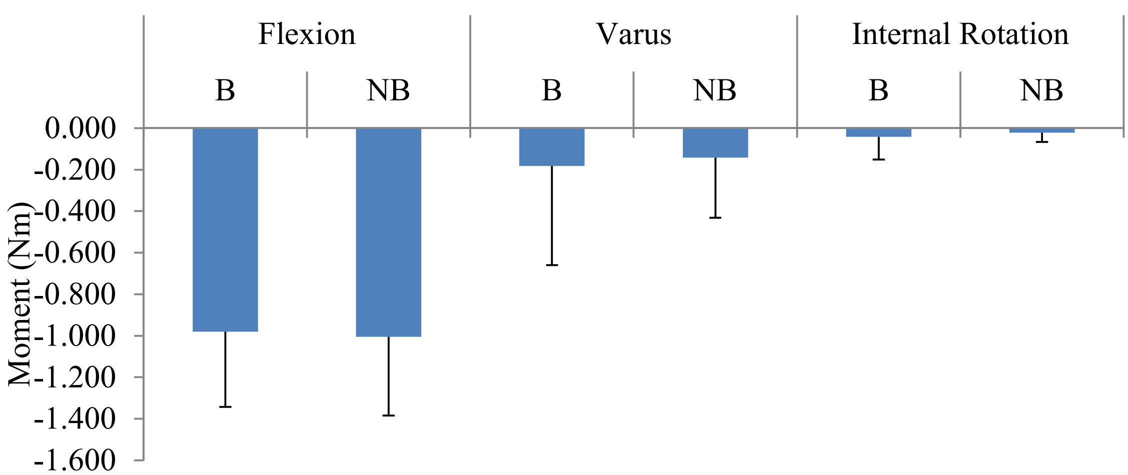

Figure 4. Brace vs no brace averages for IC knee joint angles; no significant differences found.

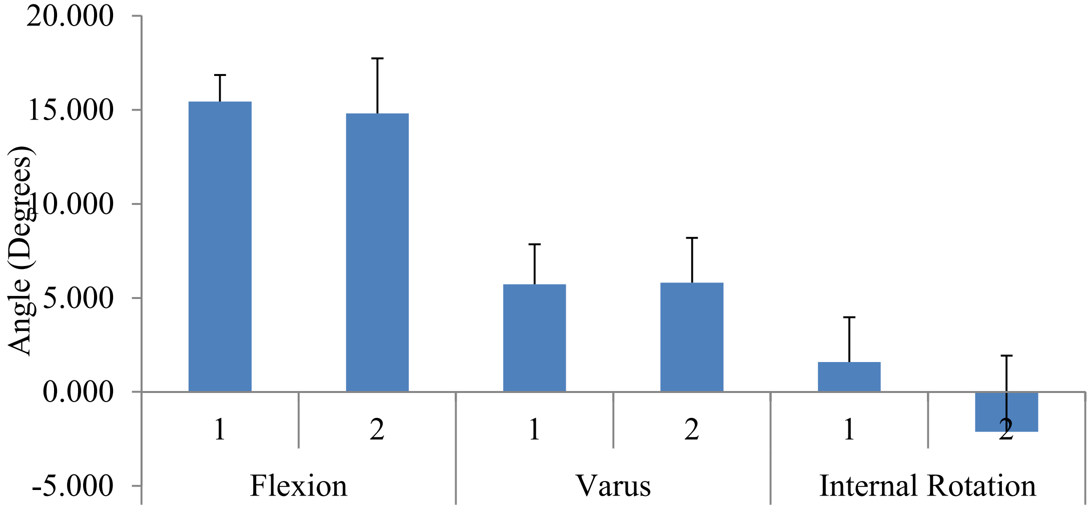

Figure 5. S1 vs S2 averages at IC for the knee joint angles; no significant differences found.

No interaction effect was found for the other two knee angles nor was any statistical difference found in comparing knee flexion/extension angle for S1 and S2 (S1: 15.44$\pm$1.41 deg; S2: 14.81$\pm$2.92 deg; $p>0.05$; see Figure 6). Additionally, no significant difference was found between B and NB for the knee flexion/extension angle (B: 13.78$\pm$1.39 deg; NB: 16.47$\pm$3.35 deg; $p>0.05$; see Figure 7). The knee internal/external rotation angle also was not significantly different between S1 and S2 (S1: 1.60$\pm$2.37 deg; S2: -2.12$\pm$4.04 deg; $p>0.05$; see Figure 7) or between B and NB (B: -0.69$\pm$3.72 deg; 0.17$\pm$2.67 deg; $p>0.05$; see Figure 7).

There was a significant order by condition interaction effect ($F=5.772$, $p=0.047$) for the knee valgus/varus angle, but there were no significant differences found in the post hoc analysis. however no significant differences were found in comparing these angles in S1 and S2 (S1: 5.73$\pm$2.12 deg; S2: 5.81$\pm$2.39 deg; $p>0.05$; see Figure 5) or these angles in B and NB (B: 4.84$\pm$2.08 deg; NB: 6.69$\pm$2.73 deg; $p>0.05$; see Figure 5).

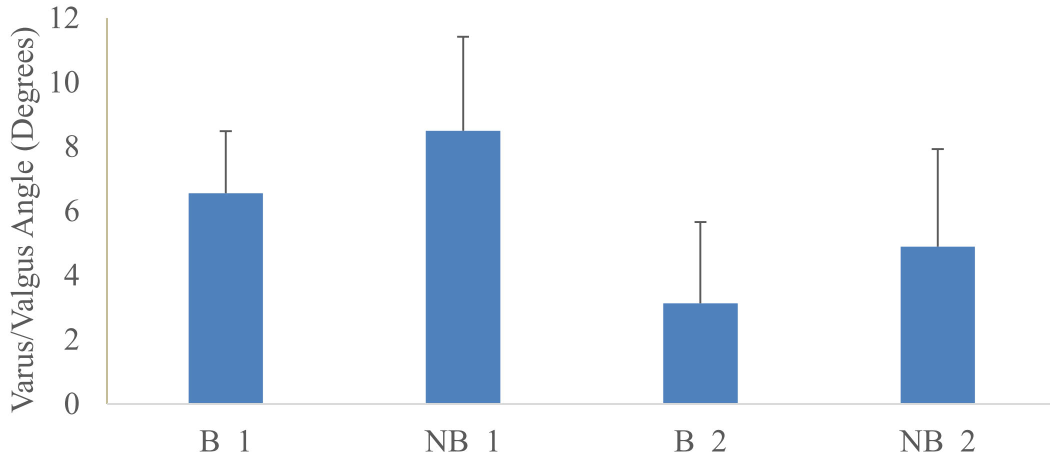

Figure 6. Post hoc analysis showed no significant differencees between brace and order conditions

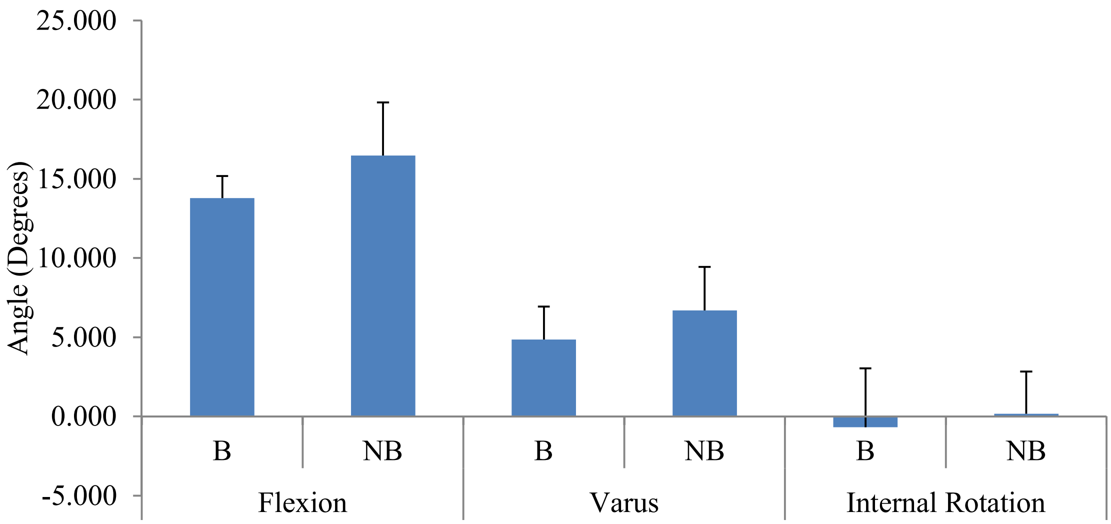

Figure 7. Brace vs no brace averages for IC knee joint angles; no significant differences found.

4.2 Moments and Angles at Maximum Vertical Ground Reaction Force

In comparing order, no statistical differences were found in the hip flexion/extension moments (S1: -0.058$\pm$0.69 Nm; S2:-0.19$\pm$0.65 Nm; $p>0.05$; see Figure 8), in the hip adduction/abduction moments (S1: -0.48$\pm$0.72 Nm; S2: 0.069$\pm$0.51 Nm;

p>0.05; see Figure 8), or in the hip internal/external rotation moments (S1:0.17±0.13 Nm; S2: 0.22±0.12; p>0.05; see Figure 8). Analysis of the hip moments at mVGRF revealed no significant interaction effects.

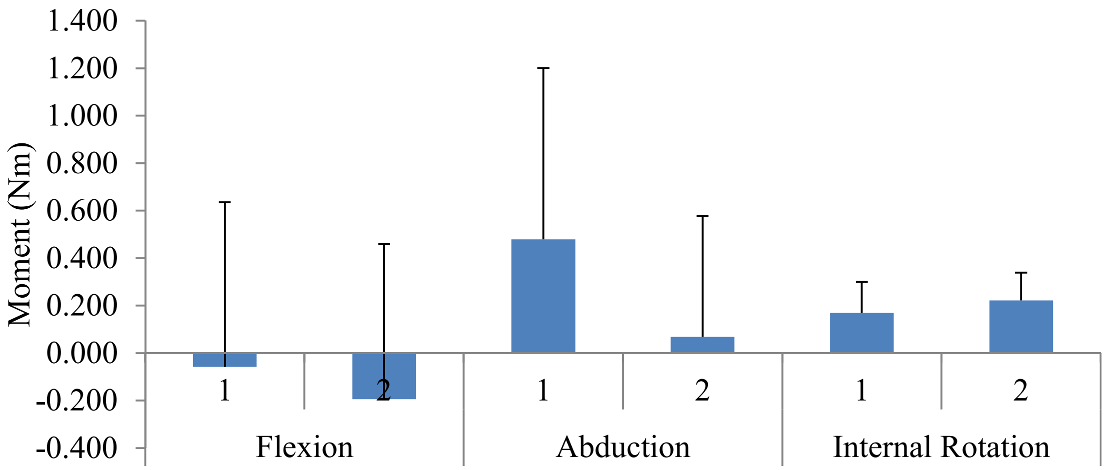

Figure 8. S1 vs S2 averages for mVGRF hip moments; not significant differences were found.

Analysis between B and NB conditions showed no significant differences for the hip extension/flexion moments (B: 0.014±0.62 Nm; NB: -0.27±0.66 Nm; p>0.05; see Figure 9), hip adduction/abduction moments (B: 0.47±0.70 Nm; NB: 0.076±0.52 Nm; see Figure 9), and the hip external/internal rotation moments (B: 0.21±0.12 Nm; NB: 0.19±0.13 Nm; p>0.05; see Figure 9).

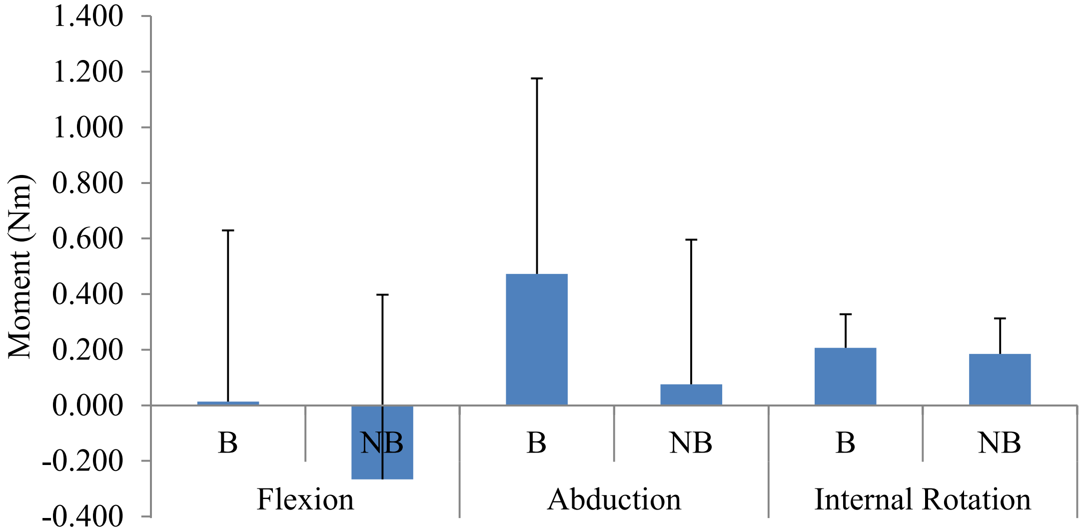

Figure 9. Brace vs no brace for mVGRF hip moments; no statistical differences

The knee varus/valgus moment showed a significant decrease ($p=0.038$) from S1 ($0.55\underline{+}0.41$ Nm) to S2 ($0.094\underline{+}0.35$ Nm), ($p<0.038$);see Figure 10).

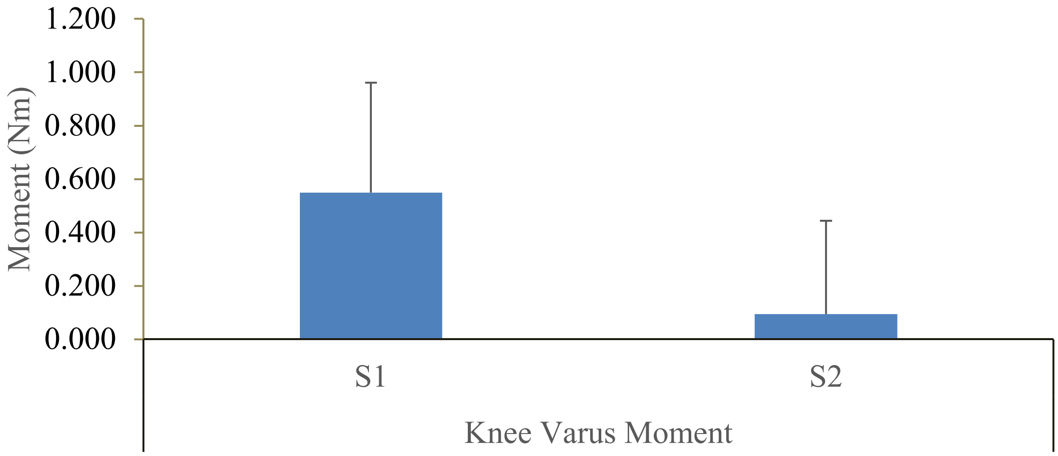

Figure 10. Comparing S1 and S2 showed a significant decrease in the knee joint varus moment.

However, no statistical differences were found for the knee flexion/extension moments between S1 and S2 (S1: $0.52\underline{+}0.31$; S2: $0.40\underline{+}0.68$ Nm; $p>0.05$; see Figure 11) nor the B and NB conditions (B: $0.089\underline{+}0.24$ Nm; NB: $0.83\underline{+}0.74$ Nm; $p>0.05$; see Figure

12). The knee external/internal rotation moments also showed no significant differences among the order conditions (S1 0.19$\pm$0.13 Nm; S2: 0.098$\pm$0.12 Nm; p>0.05; see Figure 11) and the brace conditions (B: 0.2$\pm$0.13 Nm; NB: 0.087$\pm$0.12 Nm; p>0.05; see Figure 12).

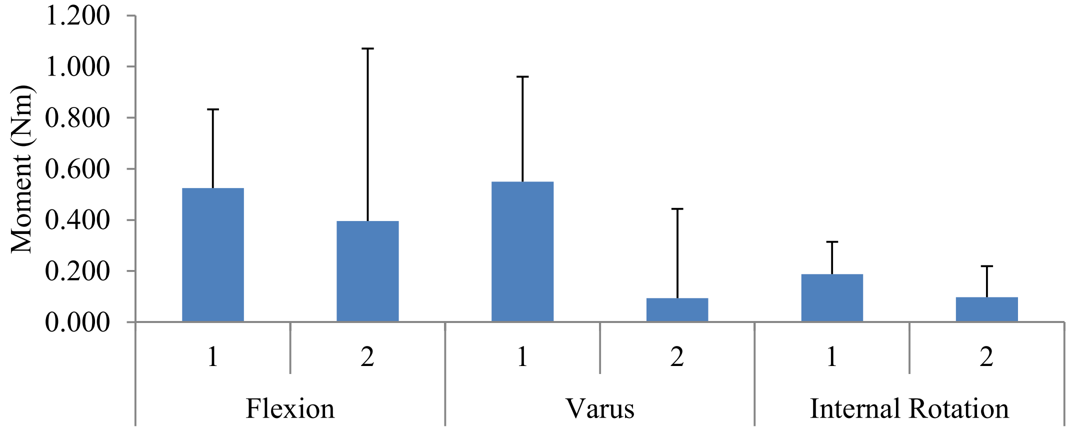

Figure 11. Varus knee moment at mVGRF was significantly lower in session 2 than in session 1; no other session difference was significant.

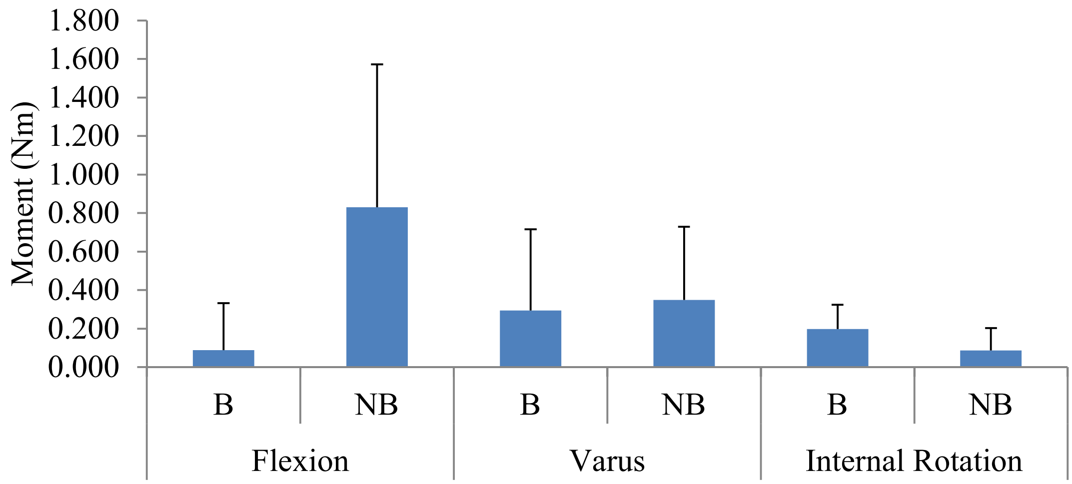

Figure 12. Brace vs no brace averages for mVGRF knee moments; no statistical difference found.

A significant interaction effect was found for the knee valgus/varus angles (F=10.341; p=0.015). Post hoc analysis showed no difference between the S1 and S2 conditions (S1: 8.75±2.21 deg; S2: 7.83±2.07 deg; $p>0.05$; see Figure 13) nor between the B and NB conditions (B: 8.25±2.15 deg; NB: 8.33±2.54 deg; $p>0.05$; see Figure 13). No significant order differences were revealed for the knee flexion/extension angles (S1: 25.02±2.16 deg; S2: 22.97±2.90 deg; $p>0.05$; see Figure 14) nor brace differences (B: 22.53±2.29 deg; NB: 25.46±2.82 deg; $p>0.05$; see Figure 15). The same result was found for the knee internal/external rotation angles for order (S1:8.79±2.89 deg; S2: 4.15±3.24 deg; $p>0.05$; see Figure 14) and brace conditions (B: 6.15±3.57; NB: 6.79±2.47 deg; $p>0.05$; see Figure 15).

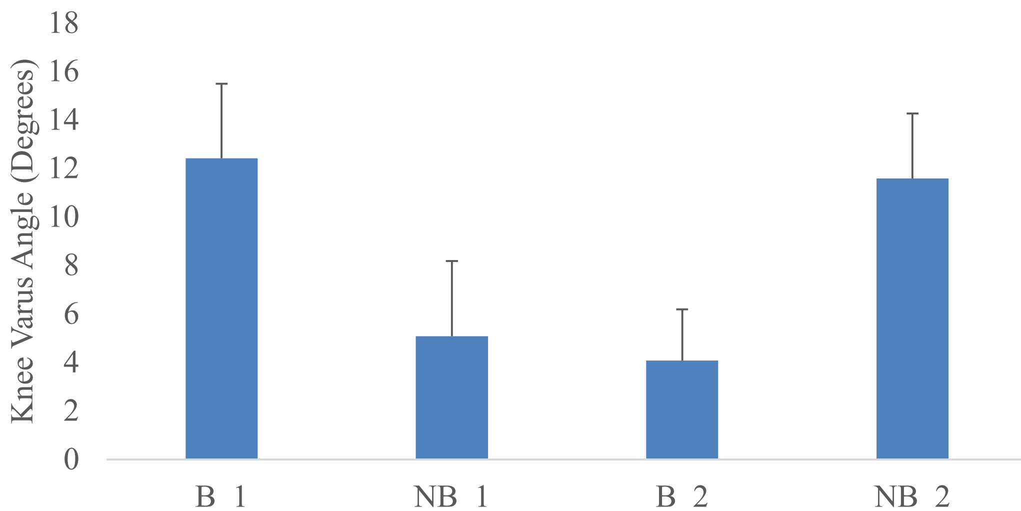

Figure 13. Brace and Order averages for the knee valgus/varus joint angle. A significant difference was found only between B_2 and NB_2 ($p=0.0095$).

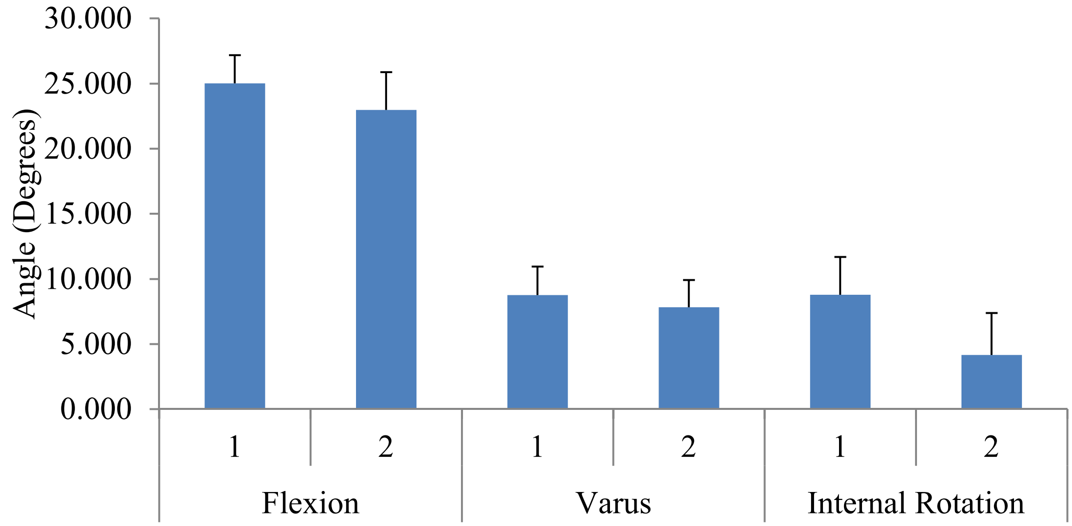

Figure 14. S1 vs S2 averages for mVGRF knee joint angles; no significant differences found.

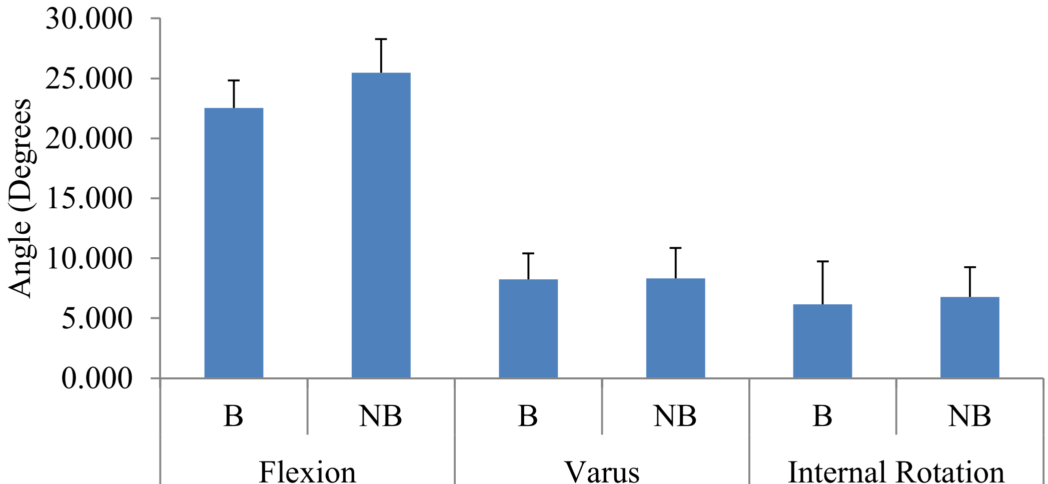

Figure 16. Brace vs no brace averages for knee angles at mVGRF; no

4.3 Moments and Angles at Maximum Knee Flexion

Evaluation of the hip moments at the MKF condition revealed no statistical significance for the brace conditions and the order of testing for the hip moments. For the order of testing, the hip flexion/extension moments showed high variability relative to the means (S1: 0.006±0.73 Nm; S2: -0.14±0.95 Nm; see Figure 16) which was also found for the hip adduction/abduction moments (S1: 0.50±0.71 Nm; S2: 0.044±0.373 Nm; see Figure 16), and the hip internal/external rotation moments (S1: 0.13±0.25 Nm; S2: 0.23±0.24 Nm; see Figure 16).

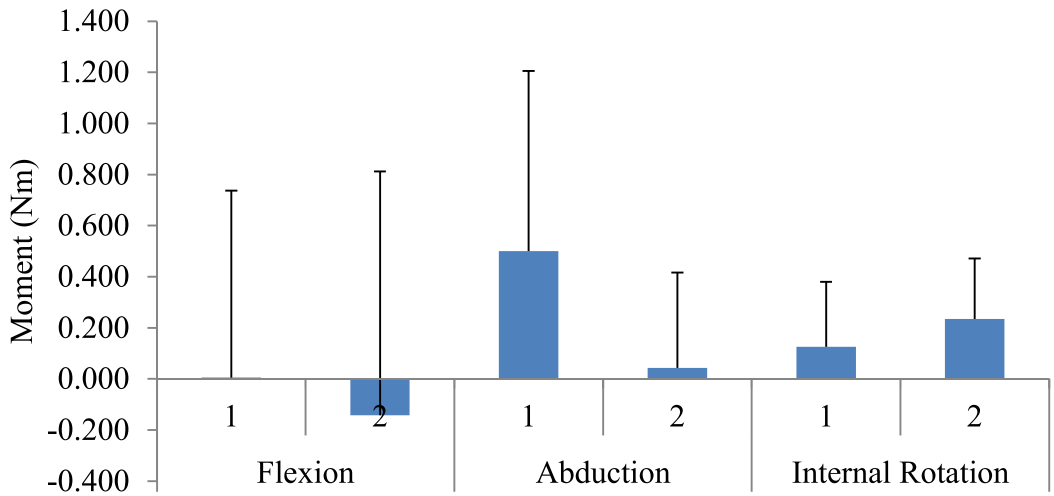

Figure 16. S1 vs S2 averages for the hip joint moment at mKF; no significant differences were found.

This feature was replicated for the brace conditions as well for the hip extension/flexion moments (B: -0.18±0.78 Nm; NB: 0.041±0.89 Nm; see Figure 17), the hip adduction/abduction moments (B: 0.29±0.68 Nm; NB: 0.26±0.34 Nm; see Figure 17), and the hip internal/external rotation moments (B: 0.25±0.26; NB: 0.11±0.23 Nm; see Figure 17).

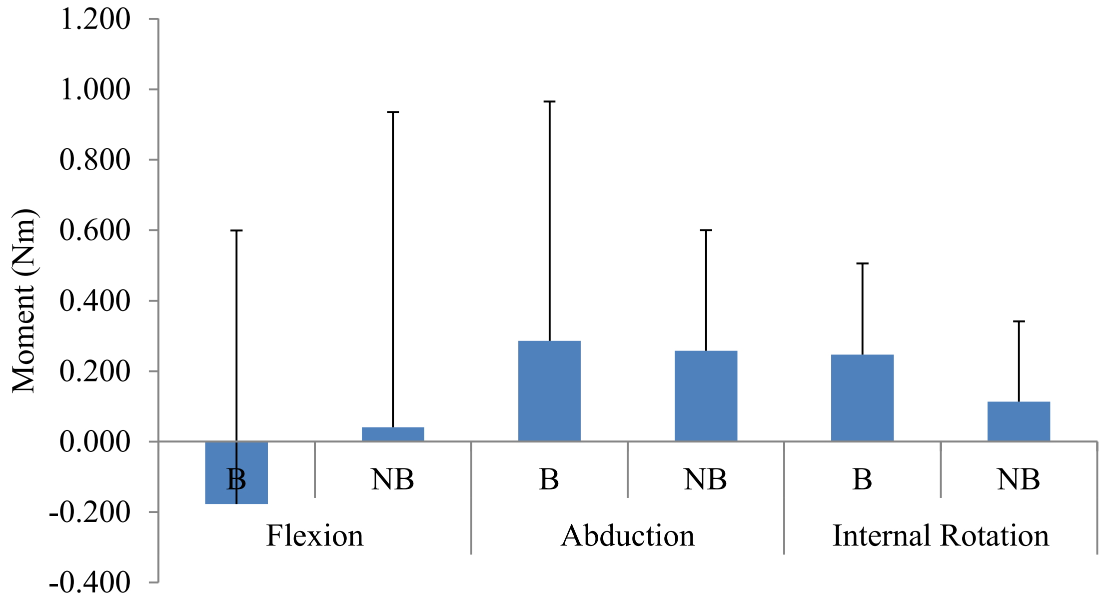

Figure 17. Brace vs no brace condition averages at mKF for hip moments; no statistical difference in the hip moments.

In addition, the knee moments had large variability and no statistical difference was found in the brace conditions and the order of testing for the knee flexion/extension moments (S1: 1.68±0.78 Nm; S2: 1.52+0.86 Nm; p>0.05; ; see Figure 18; B: 1.67±0.69 Nm; NB: 1.53±0.97 Nm; p>0.05; see Figure 19) , the knee varus/valgus moments (S1: 1.11±0.41 Nm; S2: 0.75±0.29 Nm; p>0.05; see Figure 18; B: 1.00±0.35 Nm; NB: 0.85±0.23 Nm; p>0.05; see Figure 19), and the knee internal/external rotation moments (S1: 0.26±0.26 Nm; S2:-0.029±0.16 Nm; p>0.05; see Figure 18; B:0.22±0.26 Nm; NB: 0.011±0.13 Nm; p>0.05; see Figure 19).

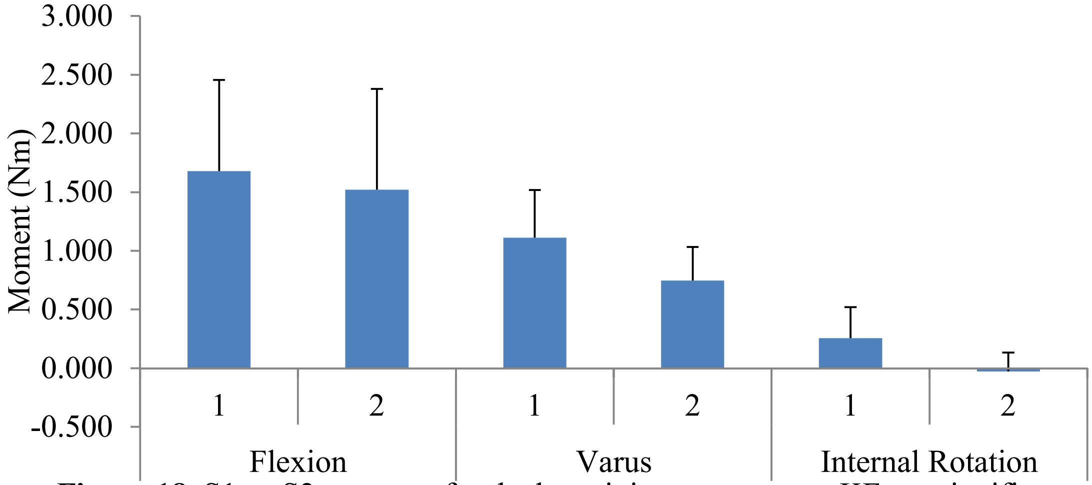

Figure 18. S1 vs S2 averages for the knee joint moment at mKF; no significant differences were found.

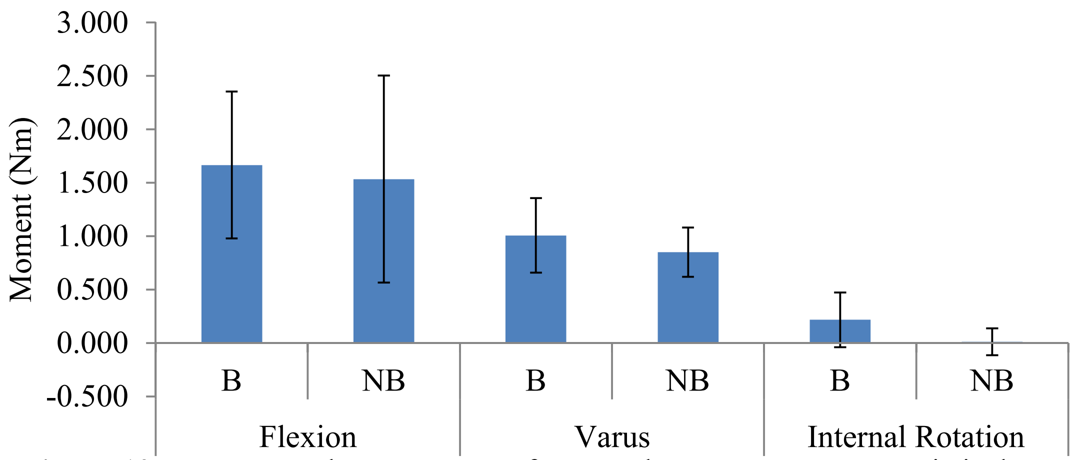

Figure 19. Brace vs no brace averages for mKF knee moments; no statistical difference was found between the knee moments.

Knee flexion/extension angles did not differ in the order of testing (S1: 53.22±2.21 deg; S2: 51.46±4.37 deg; $p>0.05$; see Figure 20) nor in the braced conditions (B: 53.36+2.47 deg; NB: 51.34±4.07; $p>0.05$; see Figure 21). The knee varus/valgus angles also showed no significant differences in the order of testing (S1: 13.27±3.41 deg; S2: 12.51±2.69 deg; see Figure 20) and in the braced conditions (B: 13.16±3.21 deg; NB: 12.62±3.04 deg; see Figure 21). Knee internal/external rotation angles also did not differ significantly between the order of testing (S1: 21.89±2.36 deg; S2: 16.55±3.25 deg; p>0.05; see Figure 20) and the braced conditions (B: 19.77±3.11 deg; NB: 18.67±2.03 deg; p>0.05; see Figure 21)

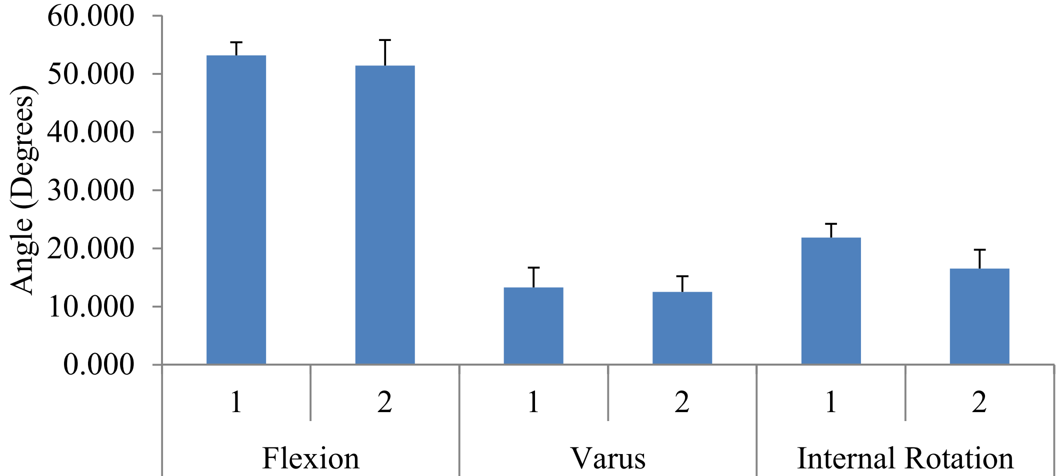

Figure 20. S1 vs S2 averages for the knee joint angles at mKF; no significant differenecs were found.

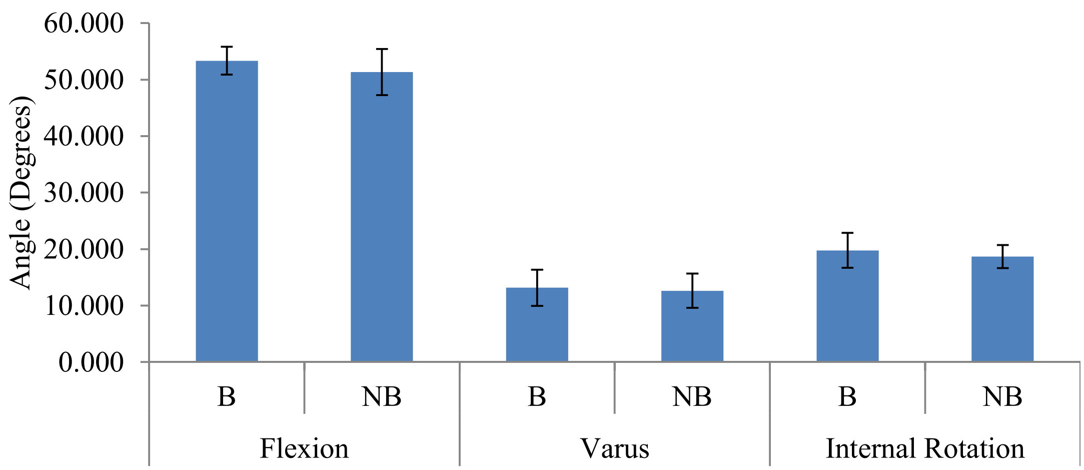

Figure 21. Brace vs no brace condition averages at mKF for knee angles; no statistical differences were found between knee angles.

Chapter 5: Discussion

The hypotheses regarding the effects of ankle bracing on knee and hip dynamics were not supported by the data. No statistical differences were found between the B and NB conditions for the knee moments, hip moments, and the knee angles. However, there was an order effect on the knee varus/valgus moment whereby from S1 to S2 there was a reduction from a varus moment to an almost zero knee varus moment. In addition, there was a significant decrease in knee varus angle from session one to session two. Stoffel et al., 2010 reported a reduction in the knee varus moment and the knee internal rotation moment when male athletes wore an ankle brace during a sidestep task like ours. Males and females have been reported to have different movement patterns during sidesteps with females having knee valgus during movements as reported by Sigard et al., 2006, Malinzak et al., 2001, and James et al., 2013. Our results do not support these reports, as on average participants performed the tasks with knee varus angles at all three time instances. A notable difference between those studies and this one is the velocity at which the movements were performed. All trials in this experiment were on average performed at 3.67 m/s while other studies had their tasks performed between 5.00 and 6 m/s. These higher running speeds were originally intended for this study but, perhaps due to the confines of the laboratory, these speeds were not attainable by these participants without anticipating potential injury risk.

One possible explanation for the reduced knee varus moment and angle in the second session could be a learning effect, as participants may have adapted their movement pattern to become more comfortable with performing the sidestep task. While recreationally active participants were recruited, a few reported they had never performed a sidestep like the one tested until this experiment and some had never worn an ankle brace before. The ankle brace acting as a catalyst could have caused a spontaneous change in motor coordination; changing from an initial motor program to another the participant was more comfortable performing. However, since half of the participants performed their first session with an ankle brace, it may have contributed to a learning effect that reduced the knee varus moment overall across both treatment orders.

The findings in this study indicate wearing an ankle brace did not increase the risk factors for injury during this sidestepping task, since wearing an ankle brace did not change the knee's angular position nor the hip and knee moments. Ankle moments were not measured in this study, as when participants wear an ankle brace it is unknown how the external, intersegmental, muscle, and brace forces interact to determine the net joint moments and forces.

Chapter 6: Conclusions

Wearing an ankle brace did not appear to have an effect on the knee and hip kinetics or the absolute knee angular position. Thus there is no apparent difference in consequences between wearing or not wearing ankle brace for the ipsilateral knee or hip. Our results indicate little or no consequences for the knee when the ankle's ROM is reduced by a standard ankle brace.

Further research is necessary to examine men and women performing the same tasks with faster approach speeds than this study. Additional research should include crossover cuts (as another direction of diagonal cutting sidestep) as well as electromyographic (EMG) measures of muscle activation timing and contraction intensity, and more kinematics of the lower limb such as angular velocities of the knee and hips. Another suggestion is to analyze the lower limb kinetics and kinematics throughout the stance phase instead of only at certain phases, so the whole movement during loading of the leg can be evaluated.

Appendix A Tables

Table 1. SPSS output for IC

Tests of Within-Subjects Contrasts											
Source				Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared	Noncent. Parameter	Observed Power[a]
Brace	KneeMomX	Linear		0.005	1	0.005	0.008	0.930	0.001	0.008	0.051
	KneeMomZ	Linear		0.003	1	0.003	0.069	0.800	0.010	0.069	0.056
	KneeAngleX	Linear		57.776	1	57.776	0.639	0.450	0.084	0.639	0.107
	KneeAngleY	Linear		27.438	1	27.438	0.638	0.451	0.084	0.638	0.107
	KneeMomY	Linear		0.013	1	0.013	0.024	0.882	0.003	0.024	0.052
	KneeAngleZ	Linear		5.794	1	5.794	0.044	0.841	0.006	0.044	0.054
	HipMomX	Linear		0.064	1	0.064	0.028	0.871	0.004	0.028	0.052
	HipMomY	Linear		0.407	1	0.407	0.214	0.658	0.030	0.214	0.069
	HipMomZ	Linear		#######	1	#######	0.002	0.964	0.000	0.002	0.050
Error(Brace)	KneeMomX	Linear		4.150	7	0.593					
	KneeMomZ	Linear		0.325	7	0.046					
	KneeAngleX	Linear		632.992	7	90.427					
	KneeAngleY	Linear		301.045	7	43.006					
	KneeMomY	Linear		3.878	7	0.554					
	KneeAngleZ	Linear		930.510	7	132.930					
	HipMomX	Linear		15.845	7	2.264					
	HipMomY	Linear		13.315	7	1.902					
	HipMomZ	Linear		0.173	7	0.025					
Order	KneeMomX		Linear	0.000	1	0.000	0.000	0.987	0.000	0.000	0.050
	KneeMomZ		Linear	0.012	1	0.012	0.682	0.436	0.089	0.682	0.111
	KneeAngleX		Linear	3.221	1	3.221	0.067	0.804	0.009	0.067	0.056
	KneeAngleY		Linear	0.060	1	0.060	0.003	0.955	0.000	0.003	0.050
	KneeMomY		Linear	0.143	1	0.143	0.414	0.540	0.056	0.414	0.087
	KneeAngleZ		Linear	110.184	1	110.184	0.744	0.417	0.096	0.744	0.117
	HipMomX		Linear	0.374	1	0.374	0.134	0.726	0.019	0.134	0.062
	HipMomY		Linear	2.760	1	2.760	2.399	0.165	0.255	2.399	0.269
	HipMomZ		Linear	0.055	1	0.055	1.355	0.283	0.162	1.355	0.173
Error(Order)	KneeMomX		Linear	4.126	7	0.589					
	KneeMomZ		Linear	0.127	7	0.018					
	KneeAngleX		Linear	339.064	7	48.438					
	KneeAngleY		Linear	121.374	7	17.339					
	KneeMomY		Linear	2.417	7	0.345					
	KneeAngleZ		Linear	1036.111	7	148.016					
	HipMomX		Linear	19.616	7	2.802					
	HipMomY		Linear	8.052	7	1.150					
	HipMomZ		Linear	0.287	7	0.041					
Brace * Order	KneeMomX	Linear	Linear	0.070	1	0.070	0.019	0.895	0.003	0.019	0.052
	KneeMomZ	Linear	Linear	0.021	1	0.021	0.108	0.752	0.015	0.108	0.059
	KneeAngleX	Linear	Linear	30.773	1	30.773	0.198	0.670	0.028	0.198	0.067
	KneeAngleY	Linear	Linear	98.830	1	98.830	5.772	0.047	0.452	5.772	0.544
	KneeMomY	Linear	Linear	0.040	1	0.040	0.010	0.922	0.001	0.010	0.051
	KneeAngleZ	Linear	Linear	154.444	1	154.444	1.074	0.335	0.133	1.074	0.147
	HipMomX	Linear	Linear	1.428	1	1.428	0.110	0.749	0.016	0.110	0.060
	HipMomY	Linear	Linear	1.828	1	1.828	0.196	0.671	0.027	0.196	0.067
	HipMomZ	Linear	Linear	0.029	1	0.029	0.779	0.407	0.100	0.779	0.120
Error(Brace*Order)	KneeMomX	Linear	Linear	26.259	7	3.751					
	KneeMomZ	Linear	Linear	1.379	7	0.197					
	KneeAngleX	Linear	Linear	1087.118	7	155.303					
	KneeAngleY	Linear	Linear	119.856	7	17.122					
	KneeMomY	Linear	Linear	27.399	7	3.914					
	KneeAngleZ	Linear	Linear	1006.990	7	143.856					
	HipMomX	Linear	Linear	90.565	7	12.938					
	HipMomY	Linear	Linear	65.144	7	9.306					
	HipMomZ	Linear	Linear	0.257	7	0.037					
a. Computed using alpha = .05											

Table 2. Averages for Order at IC

Order

Estimates

Measure		Mean	Std. Error	95% Confidence Interval Lower Bound	95% Confidence Interval Upper Bound
KneeMomX	1	-0.990	0.343	-1.801	-0.179
	2	-0.995	0.396	-1.932	-0.058
KneeMomZ	1	-0.012	0.095	-0.237	0.213
	2	-0.052	0.055	-0.182	0.079
KneeAngleX	1	15.446	1.409	12.114	18.777
	2	14.811	2.927	7.890	21.732
KneeAngleY	1	5.725	2.122	0.706	10.743
	2	5.811	2.385	0.172	11.450
KneeMomY	1	-0.095	0.413	-1.071	0.881
	2	-0.229	0.359	-1.077	0.619
KneeAngleZ	1	1.596	2.372	-4.014	7.206
	2	-2.115	4.036	-11.658	7.428
HipMomX	1	2.311	0.702	0.650	3.971
	2	2.527	0.725	0.813	4.241
HipMomY	1	-0.206	0.636	-1.710	1.298
	2	-0.794	0.342	-1.602	0.014
HipMomZ	1	-0.043	0.054	-0.171	0.085
	2	0.040	0.062	-0.107	0.188

Pairwise Comparisons

Measure			Mean Difference (I-J)	Std. Error	Sig.a	Differencea Lower Bound	Differencea Upper Bound
KneeMomX	1	2	0.005	0.271	0.987	-0.637	0.647
	2	1	-0.005	0.271	0.987	-0.647	0.637
KneeMomZ	1	2	0.039	0.048	0.436	-0.073	0.152
	2	1	-0.039	0.048	0.436	-0.152	0.073
KneeAngleX	1	2	0.635	2.461	0.804	-5.184	6.453
	2	1	-0.635	2.461	0.804	-6.453	5.184
KneeAngleY	1	2	-0.086	1.472	0.955	-3.568	3.395
	2	1	0.086	1.472	0.955	-3.395	3.568
KneeMomY	1	2	0.134	0.208	0.540	-0.358	0.625
	2	1	-0.134	0.208	0.540	-0.625	0.358
KneeAngleZ	1	2	3.711	4.301	0.417	-6.460	13.882
	2	1	-3.711	4.301	0.417	-13.882	6.460
HipMomX	1	2	-0.216	0.592	0.726	-1.616	1.183
	2	1	0.216	0.592	0.726	-1.183	1.616
HipMomY	1	2	0.587	0.379	0.165	-0.309	1.484
	2	1	-0.587	0.379	0.165	-1.484	0.309
HipMomZ	1	2	-0.083	0.072	0.283	-0.252	0.086
	2	1	0.083	0.072	0.283	-0.086	0.252

Based on estimated marginal means

a. Adjustment for multiple comparisons: Bonferroni.

Table 3. Averages for Brace at IC

Brace

Estimates

Measure		Mean	Std. Error	95% Confidence Interval Lower Bound	95% Confidence Interval Upper Bound
KneeMomX	1	-0.980	0.362	-1.837	-0.124
	2	-1.005	0.379	-1.902	-0.108
KneeMomZ	1	-0.042	0.109	-0.300	0.216
	2	-0.022	0.044	-0.126	0.083
KneeAngleX	1	13.785	1.392	10.494	17.076
	2	16.472	3.352	8.545	24.399
KneeAngleY	1	4.842	2.080	-0.077	9.761
	2	6.694	2.733	0.232	13.156
KneeMomY	1	-0.182	0.477	-1.311	0.947
	2	-0.142	0.290	-0.827	0.544
KneeAngleZ	1	-0.685	3.719	-9.479	8.109
	2	0.166	2.672	-6.153	6.485
HipMomX	1	2.374	0.731	0.646	4.102
	2	2.463	0.671	0.876	4.051
HipMomY	1	-0.387	0.712	-2.070	1.296
	2	-0.613	0.249	-1.200	-0.025
HipMomZ	1	-0.003	0.065	-0.157	0.152
	2	1.134E-05	0.039	-0.092	0.092

Pairwise Comparisons

Measure			Mean Difference (I-J)	Std. Error	Sig.a	Differencea Lower Bound	Differencea Upper Bound
KneeMomX	1	2	0.025	0.272	0.930	-0.619	0.669
	2	1	-0.025	0.272	0.930	-0.669	0.619
KneeMomZ	1	2	-0.020	0.076	0.800	-0.200	0.160
	2	1	0.020	0.076	0.800	-0.160	0.200
KneeAngleX	1	2	-2.687	3.362	0.450	-10.637	5.263
	2	1	2.687	3.362	0.450	-5.263	10.637
KneeAngleY	1	2	-1.852	2.319	0.451	-7.335	3.631
	2	1	1.852	2.319	0.451	-3.631	7.335
KneeMomY	1	2	-0.040	0.263	0.882	-0.663	0.582
	2	1	0.040	0.263	0.882	-0.582	0.663
KneeAngleZ	1	2	-0.851	4.076	0.841	-10.490	8.788
	2	1	0.851	4.076	0.841	-8.788	10.490
HipMomX	1	2	-0.089	0.532	0.871	-1.347	1.168
	2	1	0.089	0.532	0.871	-1.168	1.347
HipMomY	1	2	0.226	0.488	0.658	-0.927	1.379
	2	1	-0.226	0.488	0.658	-1.379	0.927
HipMomZ	1	2	-0.003	0.056	0.964	-0.134	0.129
	2	1	0.003	0.056	0.964	-0.129	0.134

Based on estimated marginal means

a. Adjustment for multiple comparisons: Bonferroni.

Table 4. SPSS outputs for mVGRF

Tests of Within-Subjects Contrasts

Source				Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared	Noncent. Parameter	Observed Power[a]
Brace	KneeMomX	Linear		4.405	1	4.405	1.649	0.240	0.191	1.649	0.200
	KneeMomZ	Linear		0.098	1	0.098	2.831	0.136	0.288	2.831	0.307
	KneeAngleX	Linear		68.525	1	68.525	0.913	0.371	0.115	0.913	0.132
	KneeAngleY	Linear		0.049	1	0.049	0.001	0.977	0.000	0.001	0.050
	KneeMomY	Linear		0.025	1	0.025	0.034	0.859	0.005	0.034	0.053
	KneeAngleZ	Linear		3.191	1	3.191	0.021	0.889	0.003	0.021	0.052
	HipMomX	Linear		0.626	1	0.626	0.166	0.696	0.023	0.166	0.065
	HipMomY	Linear		1.258	1	1.258	0.999	0.351	0.125	0.999	0.140
	HipMomZ	Linear		0.004	1	0.004	0.020	0.891	0.003	0.020	0.052
Error(Brace)	KneeMomX	Linear		18.701	7	2.672					
	KneeMomZ	Linear		0.243	7	0.035					
	KneeAngleX	Linear		525.584	7	75.083					
	KneeAngleY	Linear		393.553	7	56.222					
	KneeMomY	Linear		5.205	7	0.744					
	KneeAngleZ	Linear		1059.006	7	151.287					
	HipMomX	Linear		26.346	7	3.764					
	HipMomY	Linear		8.816	7	1.259					
	HipMomZ	Linear		1.367	7	0.195					
Order	KneeMomX		Linear	0.132	1	0.132	0.075	0.791	0.011	0.075	0.057
	KneeMomZ		Linear	0.065	1	0.065	1.093	0.331	0.135	1.093	0.148
	KneeAngleX		Linear	33.528	1	33.528	0.455	0.522	0.061	0.455	0.090
	KneeAngleY		Linear	6.682	1	6.682	0.259	0.627	0.036	0.259	0.073
	KneeMomY		Linear	1.660	1	1.660	6.487	0.038	0.481	6.487	0.592
	KneeAngleZ		Linear	171.591	1	171.591	1.137	0.322	0.140	1.137	0.152
	HipMomX		Linear	0.150	1	0.150	0.029	0.870	0.004	0.029	0.053
	HipMomY		Linear	1.354	1	1.354	0.904	0.373	0.114	0.904	0.131
	HipMomZ		Linear	0.022	1	0.022	0.118	0.742	0.017	0.118	0.060
Error(Order)	KneeMomX		Linear	12.236	7	1.748					
	KneeMomZ		Linear	0.416	7	0.059					
	KneeAngleX		Linear	515.786	7	73.684					
	KneeAngleY		Linear	180.675	7	25.811					
	KneeMomY		Linear	1.792	7	0.256					
	KneeAngleZ		Linear	1056.736	7	150.962					
	HipMomX		Linear	36.139	7	5.163					
	HipMomY		Linear	10.477	7	1.497					
	HipMomZ		Linear	1.326	7	0.189					
Brace * Order	KneeMomX	Linear	Linear	13.402	1	13.402	2.020	0.198	0.224	2.020	0.234
	KneeMomZ	Linear	Linear	0.010	1	0.010	0.017	0.900	0.002	0.017	0.051
	KneeAngleX	Linear	Linear	0.080	1	0.080	0.000	0.988	0.000	0.000	0.050
	KneeAngleY	Linear	Linear	440.979	1	440.979	10.341	0.015	0.596	10.341	0.787
	KneeMomY	Linear	Linear	0.008	1	0.008	0.002	0.969	0.000	0.002	0.050
	KneeAngleZ	Linear	Linear	1.975	1	1.975	0.008	0.930	0.001	0.008	0.051
	HipMomX	Linear	Linear	19.855	1	19.855	1.546	0.254	0.181	1.546	0.190
	HipMomY	Linear	Linear	1.148	1	1.148	0.076	0.791	0.011	0.076	0.057
	HipMomZ	Linear	Linear	1.001	1	1.001	3.543	0.102	0.336	3.543	0.370
Error(Brace*Order)	KneeMomX	Linear	Linear	46.437	7	6.634					
	KneeMomZ	Linear	Linear	4.191	7	0.599					
	KneeAngleX	Linear	Linear	2471.486	7	353.069					
	KneeAngleY	Linear	Linear	298.495	7	42.642					
	KneeMomY	Linear	Linear	37.383	7	5.340					
	KneeAngleZ	Linear	Linear	1688.814	7	241.259					
	HipMomX	Linear	Linear	89.909	7	12.844					
	HipMomY	Linear	Linear	106.406	7	15.201					
	HipMomZ	Linear	Linear	1.978	7	0.283					

a. Computed using alpha = .05

Table 5. Averages for Order at mVGRF

Order

Estimates

Measure		Mean	Std. Error	95% Confidence Interval Lower Bound	95% Confidence Interval Upper Bound
KneeMomX	1	0.524	0.308	-0.204	1.252
	2	0.396	0.675	-1.200	1.991
KneeMomZ	1	0.188	0.126	-0.111	0.486
	2	0.098	0.121	-0.188	0.384
KneeAngleX	1	25.020	2.160	19.913	30.126
	2	22.972	2.902	16.110	29.835
KneeAngleY	1	8.747	2.213	3.515	13.979
	2	7.833	2.070	2.938	12.727
KneeMomY	1	0.549	0.411	-0.423	1.522
	2	0.094	0.349	-0.732	0.920
KneeAngleZ	1	8.785	2.892	1.946	15.624
	2	4.154	3.240	-3.507	11.814
HipMomX	1	-0.058	0.693	-1.697	1.582
	2	-0.194	0.653	-1.739	1.350
HipMomY	1	0.480	0.722	-1.227	2.187
	2	0.069	0.509	-1.135	1.272
HipMomZ	1	0.169	0.131	-0.139	0.478
	2	0.222	0.117	-0.055	0.499

Pairwise Comparisons

Measure			Mean Difference (I-J)	Std. Error	Sig.[b]	Difference[b] Lower Bound	Difference[b] Upper Bound
KneeMomX	1	2	0.128	0.467	0.791	-0.977	1.234
	2	1	-0.128	0.467	0.791	-1.234	0.977
KneeMomZ	1	2	0.090	0.086	0.331	-0.114	0.294
	2	1	-0.090	0.086	0.331	-0.294	0.114
KneeAngleX	1	2	2.047	3.035	0.522	-5.129	9.224
	2	1	-2.047	3.035	0.522	-9.224	5.129
KneeAngleY	1	2	0.914	1.796	0.627	-3.333	5.161
	2	1	-0.914	1.796	0.627	-5.161	3.333
KneeMomY	1	2	.456*	0.179	0.038	0.033	0.879
	2	1	-.456*	0.179	0.038	-0.879	-0.033
KneeAngleZ	1	2	4.631	4.344	0.322	-5.641	14.903
	2	1	-4.631	4.344	0.322	-14.903	5.641
HipMomX	1	2	0.137	0.803	0.870	-1.763	2.036
	2	1	-0.137	0.803	0.870	-2.036	1.763
HipMomY	1	2	0.411	0.433	0.373	-0.611	1.434
	2	1	-0.411	0.433	0.373	-1.434	0.611
HipMomZ	1	2	-0.053	0.154	0.742	-0.417	0.311
	2	1	0.053	0.154	0.742	-0.311	0.417

Based on estimated marginal means

*. The mean difference is significant at the .05 level.

b. Adjustment for multiple comparisons: Bonferroni.

Table 6. Averages for Brace at mVGRF

Brace

Estimates

Measure		Mean	Std. Error	95% Confidence Interval Lower Bound	95% Confidence Interval Upper Bound
KneeMomX	1	0.089	0.244	-0.487	0.665
	2	0.831	0.741	-0.920	2.582
KneeMomZ	1	0.198	0.125	-0.098	0.495
	2	0.087	0.115	-0.185	0.360
KneeAngleX	1	22.533	2.288	17.123	27.942
	2	25.459	2.818	18.796	32.123
KneeAngleY	1	8.251	2.148	3.171	13.330
	2	8.329	2.543	2.316	14.342
KneeMomY	1	0.293	0.422	-0.705	1.292
	2	0.350	0.378	-0.545	1.245
KneeAngleZ	1	6.154	3.571	-2.291	14.598
	2	6.785	2.475	0.932	12.638
HipMomX	1	0.014	0.616	-1.442	1.470
	2	-0.266	0.664	-1.836	1.304
HipMomY	1	0.472	0.704	-1.192	2.137
	2	0.076	0.519	-1.152	1.304
HipMomZ	1	0.207	0.121	-0.080	0.493
	2	0.185	0.128	-0.119	0.488

Pairwise Comparisons

Measure			Mean Difference (I-J)	Std. Error	Sig.a	Differencea Lower Bound	Differencea Upper Bound
KneeMomX	1	2	-0.742	0.578	0.240	-2.109	0.624
	2	1	0.742	0.578	0.240	-0.624	2.109
KneeMomZ	1	2	0.111	0.066	0.136	-0.045	0.267
	2	1	-0.111	0.066	0.136	-0.267	0.045
KneeAngleX	1	2	-2.927	3.064	0.371	-10.171	4.317
	2	1	2.927	3.064	0.371	-4.317	10.171
KneeAngleY	1	2	-0.078	2.651	0.977	-6.347	6.190
	2	1	0.078	2.651	0.977	-6.190	6.347
KneeMomY	1	2	-0.056	0.305	0.859	-0.777	0.665
	2	1	0.056	0.305	0.859	-0.665	0.777
KneeAngleZ	1	2	-0.632	4.349	0.889	-10.914	9.651
	2	1	0.632	4.349	0.889	-9.651	10.914
HipMomX	1	2	0.280	0.686	0.696	-1.342	1.902
	2	1	-0.280	0.686	0.696	-1.902	1.342
HipMomY	1	2	0.396	0.397	0.351	-0.542	1.335
	2	1	-0.396	0.397	0.351	-1.335	0.542
HipMomZ	1	2	0.022	0.156	0.891	-0.347	0.392
	2	1	-0.022	0.156	0.891	-0.392	0.347

Based on estimated marginal means

a. Adjustment for multiple comparisons: Bonferroni.

Table 7. SPSS outputs for mKF

Tests of Within-Subjects Contrasts

Source				Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared	Noncent. Parameter	Observed Power[a]
Brace	KneeMomX	Linear		0.141	1	0.141	0.080	0.785	0.011	0.080	0.057
	KneeMomZ	Linear		0.339	1	0.339	0.574	0.473	0.076	0.574	0.101
	KneeAngleX	Linear		32.288	1	32.288	0.624	0.456	0.082	0.624	0.106
	KneeAngleY	Linear		2.274	1	2.274	0.031	0.866	0.004	0.031	0.053
	KneeMomY	Linear		0.195	1	0.195	0.441	0.528	0.059	0.441	0.089
	KneeAngleZ	Linear		9.628	1	9.628	0.087	0.776	0.012	0.087	0.058
	HipMomX	Linear		0.380	1	0.380	0.228	0.648	0.032	0.228	0.070
	HipMomY	Linear		0.006	1	0.006	0.003	0.956	0.000	0.003	0.050
	HipMomZ	Linear		0.144	1	0.144	1.928	0.208	0.216	1.928	0.225
Error(Brace)	KneeMomX	Linear		12.259	7	1.751					
	KneeMomZ	Linear		4.132	7	0.590					
	KneeAngleX	Linear		362.478	7	51.783					
	KneeAngleY	Linear		520.323	7	74.332					
	KneeMomY	Linear		3.091	7	0.442					
	KneeAngleZ	Linear		771.643	7	110.235					
	HipMomX	Linear		11.673	7	1.668					
	HipMomY	Linear		13.477	7	1.925					
	HipMomZ	Linear		0.523	7	0.075					
Order	KneeMomX		Linear	0.205	1	0.205	0.281	0.613	0.039	0.281	0.075
	KneeMomZ		Linear	0.649	1	0.649	0.803	0.400	0.103	0.803	0.122
	KneeAngleX		Linear	24.818	1	24.818	0.337	0.580	0.046	0.337	0.080
	KneeAngleY		Linear	4.589	1	4.589	0.072	0.796	0.010	0.072	0.056
	KneeMomY		Linear	1.077	1	1.077	0.683	0.436	0.089	0.683	0.111
	KneeAngleZ		Linear	227.642	1	227.642	1.545	0.254	0.181	1.545	0.190
	HipMomX		Linear	0.175	1	0.175	0.075	0.792	0.011	0.075	0.057
	HipMomY		Linear	1.670	1	1.670	0.586	0.469	0.077	0.586	0.102
	HipMomZ		Linear	0.094	1	0.094	0.937	0.365	0.118	0.937	0.134
Error(Order)	KneeMomX		Linear	5.108	7	0.730					
	KneeMomZ		Linear	5.654	7	0.808					
	KneeAngleX		Linear	516.031	7	73.719					
	KneeAngleY		Linear	445.660	7	63.666					
	KneeMomY		Linear	11.043	7	1.578					
	KneeAngleZ		Linear	1031.063	7	147.295					
	HipMomX		Linear	16.293	7	2.328					
	HipMomY		Linear	19.936	7	2.848					
	HipMomZ		Linear	0.702	7	0.100					
Brace * Order	KneeMomX	Linear	Linear	24.796	1	24.796	1.603	0.246	0.186	1.603	0.196
	KneeMomZ	Linear	Linear	0.001	1	0.001	0.002	0.969	0.000	0.002	0.050
	KneeAngleX	Linear	Linear	265.807	1	265.807	2.475	0.160	0.261	2.475	0.275
	KneeAngleY	Linear	Linear	414.382	1	414.382	1.633	0.242	0.189	1.633	0.198
	KneeMomY	Linear	Linear	4.925	1	4.925	1.591	0.248	0.185	1.591	0.194
	KneeAngleZ	Linear	Linear	27.165	1	27.165	0.198	0.669	0.028	0.198	0.067
	HipMomX	Linear	Linear	2.033	1	2.033	0.100	0.761	0.014	0.100	0.059
	HipMomY	Linear	Linear	0.343	1	0.343	0.028	0.872	0.004	0.028	0.052
	HipMomZ	Linear	Linear	1.915	1	1.915	0.910	0.372	0.115	0.910	0.132
Error(Brace*Order)	KneeMomX	Linear	Linear	108.281	7	15.469					
	KneeMomZ	Linear	Linear	5.557	7	0.794					
	KneeAngleX	Linear	Linear	751.866	7	107.409					
	KneeAngleY	Linear	Linear	1775.908	7	253.701					
	KneeMomY	Linear	Linear	21.670	7	3.096					
	KneeAngleZ	Linear	Linear	958.233	7	136.890					
	HipMomX	Linear	Linear	142.333	7	20.333					
	HipMomY	Linear	Linear	86.159	7	12.308					
	HipMomZ	Linear	Linear	14.735	7	2.105					

a. Computed using alpha = .05

Table 8. Averages for Order at mKF

Order

Estimates

Measure		Mean	Std. Error	95% Confidence Interval Lower Bound	95% Confidence Interval Upper Bound
KneeMomX	1	1.680	0.777	-0.158	3.517
	2	1.520	0.861	-0.516	3.556
KneeMomZ	1	0.256	0.263	-0.367	0.879
	2	-0.029	0.160	-0.406	0.349
KneeAngleX	1	53.221	2.213	47.987	58.455
	2	51.459	4.373	41.119	61.800
KneeAngleY	1	13.269	3.413	5.198	21.340
	2	12.512	2.686	6.161	18.862
KneeMomY	1	1.112	0.405	0.154	2.071
	2	0.745	0.286	0.069	1.422
KneeAngleZ	1	21.885	2.356	16.314	27.457
	2	16.551	3.246	8.875	24.227
HipMomX	1	0.006	0.731	-1.723	1.735
	2	-0.142	0.954	-2.398	2.114
HipMomY	1	0.501	0.705	-1.167	2.168
	2	0.044	0.373	-0.838	0.926
HipMomZ	1	0.126	0.254	-0.475	0.727
	2	0.234	0.237	-0.325	0.794

Pairwise Comparisons

Measure			Mean Difference (I-J)	Std. Error	Sig.a	Differencea Lower Bound	Differencea Upper Bound
KneeMomX	1	2	0.160	0.302	0.613	-0.554	0.874
	2	1	-0.160	0.302	0.613	-0.874	0.554
KneeMomZ	1	2	0.285	0.318	0.400	-0.467	1.036
	2	1	-0.285	0.318	0.400	-1.036	0.467
KneeAngleX	1	2	1.761	3.036	0.580	-5.417	8.939
	2	1	-1.761	3.036	0.580	-8.939	5.417
KneeAngleY	1	2	0.757	2.821	0.796	-5.913	7.428
	2	1	-0.757	2.821	0.796	-7.428	5.913
KneeMomY	1	2	0.367	0.444	0.436	-0.683	1.417
	2	1	-0.367	0.444	0.436	-1.417	0.683
KneeAngleZ	1	2	5.334	4.291	0.254	-4.812	15.481
	2	1	-5.334	4.291	0.254	-15.481	4.812
HipMomX	1	2	0.148	0.539	0.792	-1.127	1.423
	2	1	-0.148	0.539	0.792	-1.423	1.127
HipMomY	1	2	0.457	0.597	0.469	-0.954	1.868
	2	1	-0.457	0.597	0.469	-1.868	0.954
HipMomZ	1	2	-0.108	0.112	0.365	-0.373	0.156
	2	1	0.108	0.112	0.365	-0.156	0.373

Based on estimated marginal means

a. Adjustment for multiple comparisons: Bonferroni.

Table 9. Averages for Brace at mKF

Brace

Estimates

Measure		Mean	Std. Error	95% Confidence Interval Lower Bound	95% Confidence Interval Upper Bound
KneeMomX	1	1.666	0.686	0.043	3.289
	2	1.534	0.968	-0.757	3.824
KneeMomZ	1	0.217	0.256	-0.389	0.823
	2	0.011	0.125	-0.285	0.307
KneeAngleX	1	53.345	2.470	47.503	59.186
	2	51.336	4.068	41.717	60.955
KneeAngleY	1	13.157	3.208	5.572	20.741
	2	12.624	3.040	5.435	19.812
KneeMomY	1	1.007	0.349	0.181	1.833
	2	0.851	0.231	0.305	1.396
KneeAngleZ	1	19.767	3.107	12.421	27.113
	2	18.670	2.030	13.868	23.471
HipMomX	1	-0.177	0.777	-2.014	1.659
	2	0.041	0.894	-2.074	2.156
HipMomY	1	0.286	0.679	-1.319	1.892
	2	0.258	0.343	-0.552	1.069
HipMomZ	1	0.247	0.259	-0.365	0.859
	2	0.113	0.228	-0.426	0.652

Pairwise Comparisons

Measure			Mean Difference (I-J)	Std. Error	Sig.a	Differencea Lower Bound	Differencea Upper Bound
KneeMomX	1	2	0.133	0.468	0.785	-0.974	1.239
	2	1	-0.133	0.468	0.785	-1.239	0.974
KneeMomZ	1	2	0.206	0.272	0.473	-0.436	0.848
	2	1	-0.206	0.272	0.473	-0.848	0.436
KneeAngleX	1	2	2.009	2.544	0.456	-4.007	8.025
	2	1	-2.009	2.544	0.456	-8.025	4.007
KneeAngleY	1	2	0.533	3.048	0.866	-6.675	7.741
	2	1	-0.533	3.048	0.866	-7.741	6.675
KneeMomY	1	2	0.156	0.235	0.528	-0.399	0.712
	2	1	-0.156	0.235	0.528	-0.712	0.399
KneeAngleZ	1	2	1.097	3.712	0.776	-7.681	9.875
	2	1	-1.097	3.712	0.776	-9.875	7.681
HipMomX	1	2	-0.218	0.457	0.648	-1.298	0.862
	2	1	0.218	0.457	0.648	-0.862	1.298
HipMomY	1	2	0.028	0.491	0.956	-1.132	1.188
	2	1	-0.028	0.491	0.956	-1.188	1.132
HipMomZ	1	2	0.134	0.097	0.208	-0.094	0.363
	2	1	-0.134	0.097	0.208	-0.363	0.094

Based on estimated marginal means

a. Adjustment for multiple comparisons: Bonferroni.

Appendix B Literature Review

Human movement is a coordinated effort of the lower extremities in conjunction with the trunk and pelvis which represents a multifaceted, multisegemented system formulated by the kinetic chain principle (Kulas, Hortobagyi, & Devita, 2010; Zatsiorsky, 2002). This theory assumes the segments as rigid bodies have a range of motion in multiple planes (frontal and sagittal) at their joints (Zatsiorsky, 2002). Many studies have investigated the kinetics and kinematics of joints when they are restricted. For instance, external ankle supports (bracing and taping) have been shown to reduce inversion, eversion, dorsiflexion, and plantar flexion range of motion of the ankle (Grambo, 2014; Okamatsu, 2014; Wisthoff, 2014; Stoffel et al., 2010; Greene and Hillman, 1990; Quackenbush et al., 2008) and affects axial rotation (Santos et al., 2004). The kinetic chain principle suggests the lower extremities as a three segment system with two joints, altering the kinematics and kinetics of one joint will also affect the other joints that comprise the same system (Zatsiorsky, 2002). Thus, changing the biomechanics of the ankle via external ankle support can potentially alter the biomechanics of the knee, but whether the outcome has a protective or harmful effect is not clear.

Physical activity for people of all ages is important in order to maintain or improve their health. There are numerous benefits to being physically active which can include aerobic fitness, stronger bones, and better general wellbeing. However, an injury occurring during a bout of physical activity can have a person economic cost and takes time away from exercise. A study analyzing injury in high school students alone revealed 40.1% of all injuries that occurred in multiple sports were lower extremities injuries with the knee and ankle accounting for 29.1 and 40.5% respectively (Yang et al., 2005). Once an individual has been injured, they are at a higher risk of an injury reoccurring, however bracing or taping the ankle can reduce the risk of injury (Sitler et al., 1994; Surve et al., 1994).

Ankle taping in particular has been used with ankle multiple ankle injuries and while this provides extra support for the ankle, it is not well known what is happening to the knee. Use of ankle braces has been associated with increased rates of knee injury (Yang et al., 2005). Though it was discussed athletes with a history of knee injury will use knee braces more and may have a potential adverse effect on the non-braced leg. Thus knowing what happens to knee joint biomechanics when external ankle supports are used is of concern.

Ankle and Knee Mechanics during Landing and Cutting Tasks

Movements involving directional changes or stopping due to impact such as cutting and landing movements that are commonly used across a variety of sports such as football, soccer, basketball, and rugby have the greatest occurrence of non-contact ACL injuries (Arendt et al., 1999). In comparing injury rates among males and females, male athletes have been reported to have greater incidence of injury to the lower extremities, however female athletes are at least three times more at risk to injure their ACL (Yang et al., 2005; Arendt et al., 1999).

Landing. Injury factors for the ACL have been examined *in vitro* in order to understand what potential mechanical variables are related to ACL injuries. Withrow et al. (2006) simulated lower extremity impact in a jump landing on cadaver knees at an initial knee flexion ankle of 25 degrees, but the knees were free to flex when an impulse was present. The main findings were valgus moments in conjunction with knee flexion will place greater stress on the ACL than a knee flexion movement by itself.

In a similar study by Weinhold et al. (2007) evaluating cadaver knees during landing *in vitro* using gender specific loading parameters defined from Chappell et al. (2002), found females utilize a drop jump landing style that placed greater strain on the ACL than males. Limitations in ankle dorsiflexion range of motion can also induce ACL risk factors. Reducing ankle dorsiflexion in a squat or landing caused an increase in knee valgus and a decrease in knee flexion (Fong et al., 2011; Macrum et al., 2012; Okamatsu, 2014).

In agreeance to the kinetic chain principle, Devita and Skelly (1992) observed the knee and ankle work with some assistance from the hip in controlling segmental rotations in the lower extremities. This was observed when having participant land with either a soft (knee flexion greater than 90 degrees) or a stiff (knee flexion less than 90 degrees) landing strategy. When limiting knee flexion to the stiff landing, ankle and hip angular impulses increased while knee angular impulse was decreased compared to the soft landing. Additionally, similar contributions of work done were present in the hip, knee, and ankle in the soft landing (25, 37, and 37% respectively) as compared to the stiff landing (20, 31, and 50% respectively) except for the difference in the work contributed by the ankle.

Cutting. Overall, sidestepping and cutting movements will increase moments in the frontal and transverse plane while moments in the sagittal plane are no different compared to running (Besier et al., 2001). The combined load in sidestepping is composed of flexion, valgus, and internal rotation while flexion, varus, and internal rotation comprise crossover cutting, in regards to male athletes. Gender discrepancies have been reported in cutting tasks with females showing valgus moments compared to varus moments present in males (Sigward and Powers, 2006), greater ground reaction forces at maximum knee flexion with females (James et al., 2004), greater knee abduction angles and smaller flexion angles (Hewett et al., 2005; James et al., 2004). As discussed by Weinhold et al. (2007) and Withrow et al. (2006), increasing knee flexion places no additional stress on the ACL unless it is combined with increase in knee valgus. Females have been shown to perform motor tasks in way that increases knee valgus potentially can result in ACL injury during cutting or sidestepping tasks.

Menstrual Cycle on Joint Laxity. Evaluating knee kinetics and kinematics across the three different phases of the menstrual cycle in a cutting and jump-stop landing revealed greater knee join laxity during ovulation (Park et al., 2009; Deie et al., 2002). Increases in joint laxity were associated with increased levels of estrogen and may be helpful to explaining why females are at greater risk for injury to their ACL (Park et al., 2009). Thus it is possible that ankle taping will predispose female athletes to greater risk of ACL injury as according to the kinetic chain principle.

External Ankle Supports

With high occurrences of ankle injuries, externa ankle supports have been popularly used for many years with the possible benefit of reducing injury risk. It has been shown external ankle supports (taping and bracing) will reduce the range of motion in the frontal and sagittal plane and thus one could predict a decrease in performance, however research has shown little or no reduction in performance with the inclusion of external ankle supports. Studies have shown decreases in performance for select tasks such as agility in a right boomerang task, but no difference in balance or vertical jump (Ambegaonkar et al., 2011), while others have reported reduction in vertical jump (Mackean et al., 1995; Mayhew, 1972; Juvenal, 1972). Other studies have shown no reduction in those same as well as other performance measures (Greene and Hillman, 1990; Verbrugge, 1996; Quackenbush, 2008; Pienkowski et al., 1995).

Proprioception and Muscle Activation. The body's ability to attenuate shock depends on the positioning of the limbs (Devita and Skelly, 1992). In order for the body to be in the correct position, the muscles must fire to move a body segment into position. There must however, be feedback from the proprioceptors in the segment so it can be moved to the correct place. Karlsson and Andreasson (1992) measured the reaction times of the peroneus muscle after simulation of an ankle sprain with and without ankle tape in participants with and without unstable ankles. When the foot was taped, reaction times where shorter than the untaped condition in participants with unstable ankles. Taped and unstable ankles reaction times were less than normal reaction times in participants with stable ankles, and no difference was found between taped and untaped conditions for those with stable ankles. It was then concluded there is a proprioceptive advantage when ankles were braced

Foot proprioception while braced or taped has hardly been studied. Robbins et al. (1995) investigated whether blindfolded participants could perceive a difference in slopes while taped or untaped. The degree of slopes varied between 0 and 25 degrees in increments of 2.5 degrees. Proprioception of the angle of the foot was greater when participants were taped as opposed to untaped. Thus it is possible that not only does restricting the extreme ranges of motion of the ankle, but also increasing awareness of foot position is why providing external support to the ankle will reduce rates of ankle injury and footwear should be changed to supplement this. The increase in proprioception may result from the decrease range of motion when an ankle is braced or taped. The external ankle support will also resist this change in motion (Karlsson and Andreasson, 1992) and this resistance is felt by the shank or foot.

Loss of Support over Time with Non-Rigid Supports. Degradation of external ankle support during and post exercise has also been examined by numerous studies mainly showing that ankle taping and some braces will lose their ability to restrict motion in the frontal and sagittal plane over time (Wisthoff, 2014; Grambo, 2010; Greene and Hillman, 1990; Ricard et al., 2000). Wisthoff (2014) examined differences in lower extremity kinematics with ankle taping before and after thirty minutes of running. Post exercise evaluation revealed no reduction of movement restriction with taping or ASO in the frontal plane (inversion and eversion) or the sagittal plane until after twenty-five minutes of running. The semi-ridged hinged ankle orthosis condition was found not to degrade in movement restriction post exercise.

Grambo (2010) had participants perform fifteen minutes of multidirectional exercise with either heavy elastic or white tape, or barefoot evaluating pre and post exercise effects. In both the frontal and sagittal plane, the ankle range of motion was greater in the barefoot condition pre and post exercise, while for both tape conditions the post exercise range of motion was greater than pre exercise range of motion, demonstrating a degradation of both tape's ability to restrict motion.

Greene and Hillman (1990) investigated a comparison between ankle taping and a semi-rigid ankle orthosis at select time intervals during a three hour volleyball practice. Maximal losses for the ankle tape in the sagittal place occurred twenty minutes into exercise while the semi-rigid orthosis showed no degradation.

Ricard et al. (2000) had similar findings when subjects ran for. In conclusion, ankle taping will gradually lose its ability to restrict motion in the frontal and sagittal plane within ten minutes and will maximally lose this ability in approximately twenty to twenty-five minutes while certain semi-rigid ankle orthoses will not. Studies evaluating performance and safety measures should take this into consideration when measuring across multiple tasks.

Knee and Ankle Biomechanics with Ankle Support

While external ankle support may not have adverse effects on performance, there could still be an adjustment strategy made by athletes to compensate for the lost range of motion. Santos et al. (2004) examined three trunk turning tasks with and without an ankle brace: standing on one leg (right leg) and turning the trunk, turning sideways to catch a ball, or rotate the trunk to the left so the shoulder touches a target in front. With an ankle brace, participants showed reduced trunk axial rotation, but no increased in knee rotation when catching a ball. Santos et al. (2004) suggested compensation was made by the upper extremities to achieve the task. In the task to turn and touch a shoulder to a target, as a certain degree of trunk rotation was required, no compensation could be made by the trunk. Compensation was however, made by increasing internal rotation of the knee. Thus adaptation strategies are task-dependent.

Landing. As external ankle supports will reduce plantar flexion and dorsiflexion, a compensation strategy then must be made during landings. Devita and Skelly (1992) showed that the ankle joint is a major contributor to work done when different landing strategies are performed without ankle supports. Several studies have tested the use of external ankle supports in landings and the effects on the knee. Venesky et al. (2006) had participants drop from an adjustable bar onto a slanted board with their dominant leg with the ankle either braced or unbraced. The board was slanted to induce the foot to invert and to resist this motion, and ankle eversion torque must be present. Results show that the ankle eversion torque and knee external toque were greater in the brace condition than without a brace, while knee valgus torque was the same.

External torque in the knee has been suggested to increase injury risk (Santos et al. 2004). In a study examining female landing strategies with external ankle supports (brace), medial knee displacement, vertical ground reaction forces were increased while dorsiflexion and ankle displacement decreased when an external ankle support was used. Median knee displacement was defined essentially as knee valgus, and as such in conjunction with increased vertical ground reaction forces increases the risk for ACL injury (Weinhold et al., 2007; Withrow et al., 2006).

Long and Short Term Effects. Most studies have only evaluated the short term effects of ankle bracing. If a compensation strategy is being performed when a joint is restricted, then repeated trials could potentially provoke motor development in order to increase efficiency. A study by DiStefano et al. (2008) evaluated acute and chronic effects of ankle bracing over a period of eight weeks. There were no differences in wearing the brace for eight weeks. Acute effects include restrictions in the ankle's sagittal plane (dorsiflexion and plantarflexion) and knee flexion range of motion meanwhile knee flexion at contact was increased while vertical ground reaction forces did not change. It is important to note that the study did not measure frontal plane kinematics or kinetics. If a greater knee valgus or varus was found with the knee flexion with a braced ankle, this would increase the risk for injury.

Knee Biomechanics during Running and Cutting Tasks with External Ankle Supports

While cutting tasks have already been shown to have adverse effects on the knee (Besier et al., 2001; James et al., 2004; Sigward and Powers, 2006; Hewlett et al., 2005), it is not clear what happens when the ankle is braced or taped during cutting tasks to the knee. Stoffel et al. (2010) reported a protective response with ankle tape when examining rugby players perform unplanned and planned sidestepping and running. Crossover cuts were performed in the study but were not analyzed. Peak internal and varus moments, and valgus angles of the knee were reduced when the ankle was taped. It was stated by the study that semi-pro or elite rugby players were recruited to participate in the study to reduce variability in the cutting movements as they are more adept at reproducing movement. This study only examined male athletes.

In the study by Histhoff (2014), three different external ankle supports were evaluated against no support during a running task. This study only examined the kinematics of the ankle, knee, and hip. A significant decrease in the transverse plane range of motion of the knee (internal rotation) was found in the taped condition while no differences were found between the semi-rigid ankle orthosis and ankle brace. This effect lasted until twenty-five minutes into exercise due to degradation of the tape. Knee flexion-extension excursion was also decreased in the taped condition. Whether or not a protective response was found cannot be decided. While internal rotation was decreased, knee flexion-extension was also decreased. Ground reaction forces were not measured in this study; however the reduction in knee flexion-extension potentially represents an increase in vertical ground reaction forces (Devita and Skelly, 1992).

Another recent study hoped to evaluate both kinetic and kinematics of the knee in unplanned cutting tasks when the ankle is braced (Plumlee, 2011). The study recruited both men and women to participate in the study. However, several participants data were not included and was noted by the author to cause a small effect size, thus a sample size of twenty to twenty-two participants be studied. Possibly due to the small sample size, no significant effects were found between taped and untaped conditions during cutting maneuvers. Hence there were no protective benefits or harmful effects on the knee when the ankle was taped. The ankle's range of motion however was still restricted and thus contradicts the kinetic chain principle. It was suggested that future research investigate more arduous cutting maneuvers to both the dominant and non-dominant sides and examine muscle activity.

Summary

Ankle sprains are regularly occurring injuries in many sports and are commonly treated with bracing or taping of the ankle (Yang et al., 2004). Gender differences in landing and cutting tasks have been well established, but gender differences with the use of external ankle supports has not. Females have higher rates of ACL injuries and thus should be included in future studies concerning ankle bracing its effects on the knee. It appears more studies are in agreement with the kinetic chain principle (Devita and Skelly, 1992; Santos et al., 2004; Stoffel et al., 2010, Histhoff, 2014) than against it (Plumlee, 2011). Effects of ankle taping on biomechanics is clear in that there will be a reduction in both the frontal and sagittal plane range of motion and motion in those planes is resisted (Venesky et al., 2006; Wisthoff, 2014; Grambo, 2010; Karlsson and Andreasson, 1992). There seems to be a literature gap in regards to external ankle supports effects on the knee during cutting maneuvers (crossover cut and sidestep) for both males and females to this reviews knowledge as only two studies were found. (Stoffel et al., 2010 Plumlee, 2011) and only one evaluated males against females.

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Vita

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